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Review of Efforts to Develop a Low-Luminance-Level Disability Glare Tester

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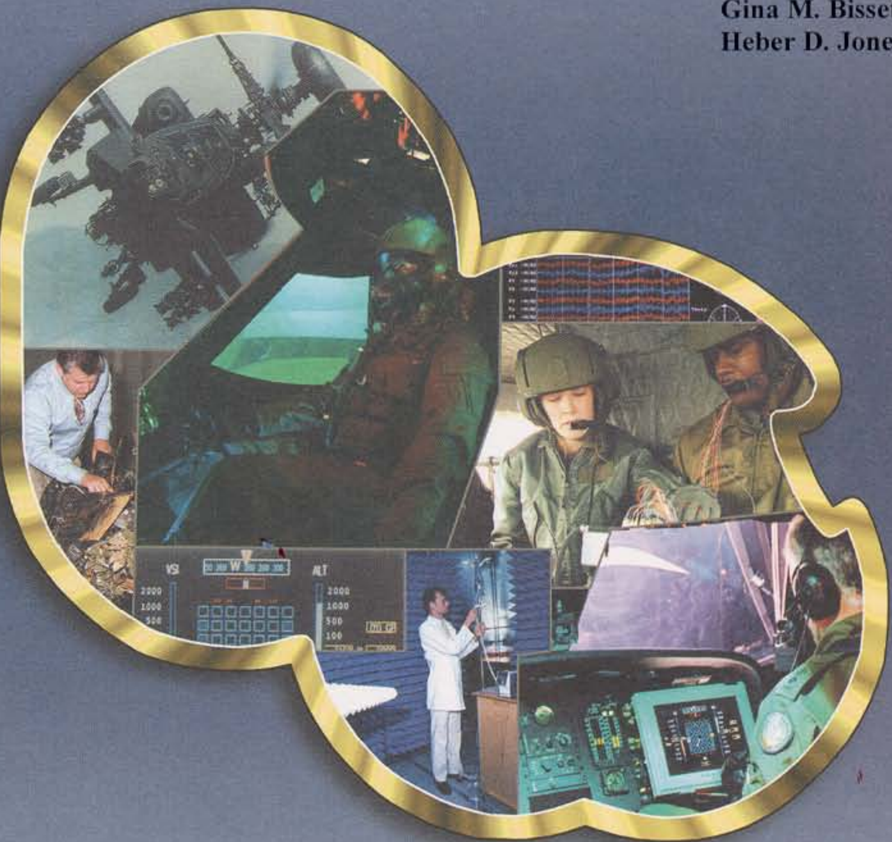

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14. ABSTRACT

This report describes USAARL's efforts to develop a disability glare tester that could be used for predicting night vision problems resulting from post refractive surgery, particularly for U. S. Army aviators. The nature & complexity of the problem are reviewed, along with current methods for assessing disability glare. The best current techniques focus on measuring & evaluating the effects of forward scatter in the eye. While elegant, forward scatter measurements alone do not reliably predict night vision problems due to aberrations induced by refractive surgery. Low-level efforts to develop a procedure that can predict night vision problems resulting from aberrations induced by refractive surgery have been pursued at USAARL for many years. The latest efforts focus on developing hybrid procedures that use an alternating spot threshold, with &d without a pupil-sparing glare annulus & pedestal, & a single spot threshold combined with a single-spot increment threshold procedure to obtain an indication of both blur & intraocular forward scatter. Further refinement & testing of these techniques is necessary to determine the efficacy of the approaches.

15. SUBJECT TERMS

glare, halo, forward scatter, aberrations, blur, disability glare, disability glare measurement, glare tester, refractive surgery, post refractive surgery night vision problem prediction and evaluation

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Table of contents

List of figures

Table of contents (continued)

List of figures

List of tables

Abstract

This report is a history and a progress report on efforts at the U.S. Army Aeromedical Research Laboratory (USAARL) to develop a disability glare tester that could be used for predicting night vision problems resulting from post refractive surgery, particularly for U. S. Army aviators. The nature and complexity of the problem are reviewed, along with current methods for assessing disability glare. The best current techniques focus on measuring and evaluating the effects of forward scatter in the eye. While elegant, forward scatter measurements alone don't reliably predict night vision problems due to aberrations induced by refractive surgery. Low-level efforts to develop a procedure that can predict night vision problems resulting from aberrations induced by refractive surgery have been pursued at USAARL for a number of years. The latest efforts have been focused on developing hybrid procedures that use an alternating spot threshold, with and without a pupil-sparing glare annulus and pedestal, and a single spot threshold combined with a single-spot increment threshold procedure to obtain an indication of both blur and intraocular forward scatter. Further refinement and testing of these techniques is necessary to determine the efficacy of the approaches.

Introduction and military significance

Disability glare reduces visual performance by reducing image contrast and/or distracting an individual visually. Disability glare is a consequence of intraocular scatter or stray light, diffraction, imperfect focus, or an inhomogeneous optical medium. It results when a peripheral light source interferes with the perfect rendering of the image of another angularly near object, as with the glare resulting from oncoming automobile headlights. Usually, but not always, glare is transient, being a serious problem only when it happens during some critical moment. It can be purposely induced or inadvertent, causing an inability to detect or identify an object on the side of the road or ahead while driving at night; a temporary inability to read instruments while flying or targeting an enemy; a reduced capability when using night vision devices. Refractive surgery often exacerbates the effects of and increases susceptibility to disability glare, consequently increasing both its magnitude and frequency of occurrence and increasing risk to self and others on the battlefield.

The ability to measure and predict who has higher risk of disability glare is inadequate at this time, particularly for post refractive surgery patients. Early LASIK and PRK produced a very high incidence of night vision problems, including glare, halos, starburst patterns and haze. More recently, with the advent of larger optical and graded transition zones, the incidence has seemingly been reduced. The use of better aberration measurement and more accurate ablation and healing strategies will reduce the problems even further. However, the problem has not gone away. The need for better disability glare measurement techniques remains. It is important to develop a broader understanding and better measurement of glare than has previously been the case. It is not sufficient to simply take contrast sensitivity measurements with a glare source or to measure intraocular stray light produced from scatter. Refractive surgery induces optical aberrations in the cornea and changes the compensating mechanisms of the eye. These are factors that must also be taken into account in the development of any new disability glare

measurement procedure. This paper explores techniques for measuring and evaluating disability glare in order to predict glare-related problems during night vision, particularly problems resulting from refractive surgery. The desired outcome of this kind of research should be a reliable, easy-to-use clinical procedure for measuring and predicting disability glare associated with night vision.

Background

Disability glare history

Glare was described by Goethe in 1810 and Purkinje in 1823. Their explanations portended the neural versus physical (light scatter) debate that was clearly framed by Helmoltz in 1852. Cobb, in 1911, was the first to quantify disability glare by developing the concept of *equivalent background* (Vos, 2003; Franssen et al, 2006). The concept was expanded by Holladay (1926, 1927), Stiles (1929), and Stiles and Crawford (1937). Their work, formally presented at the 1939 Commission Internationale de l'Eclairage (CIE) meeting, culminated in a formula that clearly implied that intraocular scatter was the main cause of glare:

$$L_{eq} = 10E_{glare} / \theta^2 \qquad (1)$$

where L_{eq} is the equivalent veiling background in cd/m^2, E_{glare} is the illuminance of the glare source at the eye measured in lux, and θ is the angular distance between the line of sight and the glare source in degrees. For an extended glare sources this formula is integrated over the angular aperture of the glare source. Subsequent research, carefully controlling pupil size and eye movement, substantiated the proportionality of L_{eq} and E_{glare}. In addition, it was shown that the forward scatter from the cornea, crystalline lens and ocular fundus, taken together, are sufficient to explain L_{eq} (Vos, 2003).

The Holladay-Stiles formula is still widely used and considered a good estimate for glare from sources between 1^o and 30^o. As a post script, it should be noted that this formula was widely used during World War II, the intervention of which probably prevented their work from becoming an early standard

Le Grand, in 1937, proposed a system for measuring forward scatter in the human eye (Thomson, 2001) that was later exploited and developed by van den Berg (1986, 1991, 1994, 1995), van den Berg and Boltjes (1988), van den Berg et al. (1989, 1991) and IJspeert et al. (1990). Their approach used a bright, flickering annulus. Forward scatter was measured by adjusting the brightness of a central disc pedestal, flickering in counter-phase to the annulus, until there was no discernable central flicker.

Fry and Alpern (1953) referenced a 1939 observation by Schouten and Ornstein, "…that the depression of brightness still persists when the image of the glare source falls on the optic nerve head," an area without receptors and lateral neural connections. Fry and Alpern found that the course of fovea dark adaptation following a peripheral glare source or a direct veiling illumination followed the same pattern. In addition they showed that increasing the glare angle was equivalent to decreasing the direct veiling illumination. These studies argued that the

brightness at the fovea from the test object was a consequence of forward light scatter in the eye caused from this peripheral glare source and not lateral neural effects.

DeMott and Boynton (1957) photographed directed light exiting excised steer eyes. This direct technique allowed them to obtain estimates of forward scatter from the cornea and lens. Rushton and Gubisch (1966) raised the fovea visual threshold using a luminous annulus, followed at another time by direct fovea light stimulation to an equivalent threshold. They then compared the central bleaching of fovea photoreceptors based on the two methods of stimulation using retinal densitometry. The cone photoreceptors were bleached equally, which is compatible with a non-neural, scatter-only hypothesis.

Around 1965, the CIE asked Vos to head a committee to update the Holladay-Stiles formula. He had recently completed a doctoral dissertation on the mechanisms of glare. In a succession of papers that followed he showed that, with some variability, the cornea, lens, and fundus contributed about equally to forward scatter in the normal eye (Vos, 1963; Vos and Boogaard, 1963; Vos and Bouman, 1964). Vos also showed that the three sources of scatter alone could account for the L_{eq} in the Holladay-Stiles formula, putting to rest the physical scatter-neural controversy.

There were other major issues regarding scatter that also had to be solved. One had to do with the question of wavelength. In general it has been found that stray light (scatter) in the eye is independent of wavelength (Wooten and Geri, 1987; van den Berg et al., 1991; Vos, 2003). However, van den Berg et al. (1991) found a small wavelength-dependent scatter with transmission of light through the ocular wall of subjects with blue eyes. The effect was virtually zero for subjects with dark brown eyes. They concluded that "depending on pigmentation, eye-wall transmittance and fundus reflections do introduce some wavelength dependence." This suggests that most scatter in the eye is Mie scatter, due to intraocular and intracellular particles substantially larger than the wavelengths of visible light. This is consistent with scatter produced by most cataracts and lens opacities resulting from hereditary factors, trauma, inflammation, UV radiation, drugs, or disease (de Waard et al., 1992; Klein, Klein, and Linton, 1992; Schneck et al., 1993; Thomson, 2001; Smith, 2002; Kanski, 2003). Cataracts are usually whitish, occasionally brunescent, and are made up of fairly large particles. The amount of scatter in the normal eye that is independent of wavelength has also been shown to be related to eye pigmentation; more pigmented eyes generally show less scatter (van den Berg et al., 1991; Vos and van den Berg, 1999; Vos, 2003).

Another major factor affecting disability glare is age (IJspeert et al., 1990; Vos and van den Berg, 1999; Vos, 2003). De Waard et al. (1992) found that light scatter increases by a factor of three by age 80. Schieber (1994a, 1994b, 1995) extensively reviewed the impact of visual aging on driving performance, pointing out that there is not only an increase in glare sensitivity with age, but also an increase in glare recovery time. Swanson (1998) pointed out that scatter increases significantly with age and that as little as 1/3 of the light reaching the retina in a 25-year-old reaches the retina of a 65-year-old. Additionally, he pointed out that light scatter in the lens is responsible for a majority of the complaints of disability glare for older adults, often leading to a voluntary cessation of night driving. Haegerstrom-Portnoy, Schneck, and Brabyn

(1999) showed that, even though everyone experiences disability glare to some extent, the effect is accelerated after age 65. Vos (2003) and Vos and van den Berg (1999) described a disability glare equation:

CIE General Disability Glare Equation, valid in the glare angle domain $0.1^o < \theta < 100^o$, reads:

$$(L_{veil} / E_{glare})_{general} = 10 / \theta^3 + (5 / \theta^2 + 0.1\, p / \theta) \times (1 + [Age / 62.5]^4) + 0.0025p \qquad (2)$$

in which θ is in degrees, L_{veil} in cd / m^2 and E_{glare} in lux. Note the switch from L_{eq} in Equation (1) [the Holladay-Stiles formula] to L_{veil} in Equation (2), reflecting the new insight that the veil is more than a computational entity and is a light veil due to entoptic scatter. One easily recognizes in Equation (2) the indicated features, such as the increased steepness on θ at small angles, the age dependency in the middle angular region, and the dependence on ocular pigmentation, p (ranging from p = 0 for very dark eyes to p = 1.2 for very light eyes) at very large glare angles.

There have been many attempts at measuring disability glare, but no universally adopted technique (i.e., no gold standard). The direct compensation technique currently gaining the widest support in Europe has adopted the stray light definition of glare and called the problem solved, i.e., glare is the veiling light that results from forward Mie scatter in the eye (De Wit, 2002, van den Berg, project leader of the Assessment of Visual Function of Driving-licence Holders, 2002, van den Berg and van Rijn, project leaders of the European Glare Project, 2005: van Rijn et al., 2005). Van Rijn, Nischler et al. (2005) stated that:

> "…disability glare is the reduction in visual performance caused by veiling luminance on the retina (illuminance at the retina). It is an effect of intraocular stray light. Measurements of glare and stray light are particularly important for [automobile] drivers, cataract [patients], and refractive surgery [patients]. Glare testing in the elderly may be important in view of the high accident rates in this age group, especially at night. Moreover, glare measurements may predict future decrease of visual acuity." (p.345)

This would be fine, except that this approach does not, for example, adequately predict the nighttime "glare" experienced by refractive surgery patients when they see the headlights of an oncoming vehicle. Stray light measurement is certainly a major factor in creating glare and works very well for evaluating cataract patients. However, with the advent of refractive surgery (figure 1) other factors have come into play. Van den Berg (1991) questioned the validity of most glare testing, stating that:

> "…present tests for the straylight type of glare fail on validation research. Also, in clinical use, the reliability of glare testing seems to be questionable. The problem is the absence of a generally accepted reference, a golden standard of glare." (p.180)

With his development of the Straylight Meter and subsequent computerized versions (discussed below) van den Berg has changed his viewpoint somewhat (van Rijn and Nischler et al., 2005), generally adopting the forward scatter-only position.

Ghaith et al (1998) found that disability glare assessments 1, 3, and 6 months post-radial keratotomy (RK) and photorefractive keratectomy (PRK), using measurements from the Brightness Acuity Tester (BAT) and the Multivision Contrast Tester (MCT 8000), did "not accurately reflect patient's subjective assessment of their visual performance in daily life as expressed in a questionnaire" (p.12). These devices measured high and low contrast visual acuity (VA), which should be affected by veiling glare resulting from forward Mie scatter. In a personal correspondence, Barbur (2004) said that visual performance of most refractive surgery patients does not differ significantly from normal subjects having had no surgery. However, he pointed out that there are some significant "outliers" that present with demonstrable vision problems, particularly when under mesopic ambient illumination, when a large pupil size favors increased aberrations. There is "a need for a simple 'pupil-sparing' visual performance test that reflects both the effects of increased scatter and aberrations."

There is a well-known paradox in glare testing that uses a bright source and low contrast acuity targets (figure 2)--visual performance can improve when the glare source is turned on (Berman et al., 1993, 1996; Nakagawara, 1994; Whitaker, Steen, and Elliott, 1994; Rabin, 1994, 1995; van den Berg, 1994, 1995; Cox, Norman, and Norman, 1999; Wachsler et al., 1999; Thomson, 2001). There are several reasons for this, most of which are related to the reduced pupil size resulting from a bright glare source, i. e., removal of a peripheral lens opacity from the pupil due to reduced pupil size, decrease in the effects of peripheral corneal aberrations by reducing pupil size, and reduction of blur due to a pinhole effect.

Using analysis of videokeratographs, Martinez et al. (1998) showed that:

> "…opening the pupil from 3 to 7 mm increased the spherical-like aberrations 7-fold before PRK. After PRK, however, pupillary dilation caused a 300-fold increase in this type of aberration…The problem with devices that measure glare visual acuity is that they cause pupillary constriction, creating a pupil that is small (creating optics close to the diffraction limit), and they provide little information about how the junction between the treated and untreated cornea affects vision." (p.1053 & p.1060)

They went on to say that the total aberrations of the cornea usually decrease to pre-surgery levels for a 3 mm pupil some months after surgery. However, the increased total aberrations created by surgery are probably permanent for larger pupil diameters. Seiler et al. (2000) measured corneal aberrations following PRK (6-7 mm ablation zones) with a Tscherning type aberroscope, reporting results as Zernike coefficients. They found that the average total wavefront error increased by a factor of 17.64. They also pointed out that aberrations are the main cause of reduced visual performance following refractive surgery. This work was corroborated by Oshika et al. (1999). They found that before PRK or LASER in-situ keratomileusis (LASIK), simulated pupil dilation from 3 to 7 mm caused a 5-6 fold increase in the total aberrations. "After surgery, the same dilation resulted in a 25- to 32-fold increase in the PRK group and a 28- to 46-fold increase in the LASIK group. The coma-like aberration was significantly increased after surgery and remained at high level throughout the one-year study period." (p.6)

The major post refractive surgery visual problems that patients experience are night vision glare, reduced contrast sensitivity, halos and starburst (Fan-Paul et al., 2002; Bailey et al., 2003). These problems are usually reduced a few months after surgery (McLeod, 2001). However, it is not entirely clear whether this is a resolution of the problem, patient adaptation, or simply self-justification, i. e., resolution of cognitive dissonance (Brunette et al., 2000; Chou and Wachler, 2001; Melki, Proano, and Azar, 2003).

LASIK **PRK**

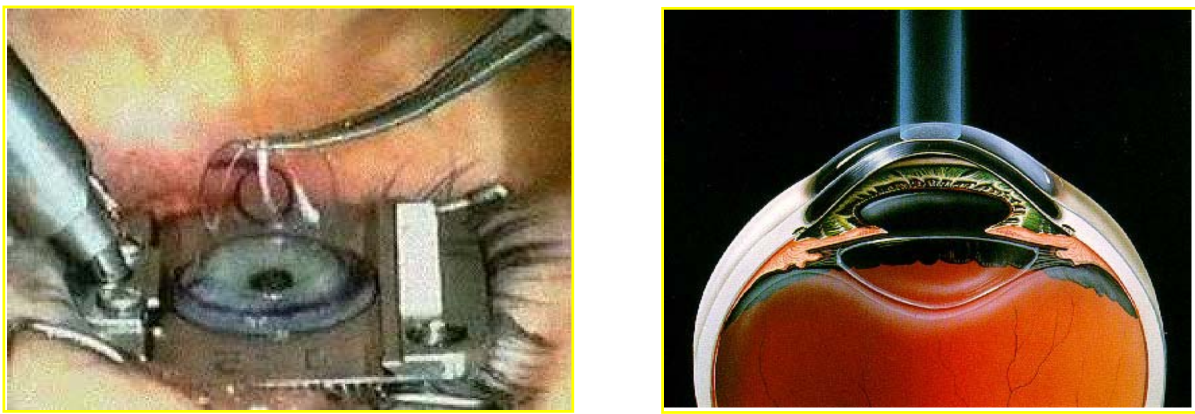


Figure 1. LASIK and PRK refractive surgery. LASIK refractive surgery removes a flap from the cornea before LASER ablation. After LASER ablation the flap is replaced. PRK refractive surgery removes only the top layer of cells (epithelium) from the cornea prior to LASER ablation. These cells grow back, from the periphery toward the center. After six months the visual performance outcomes of these surgeries are essentially the same.

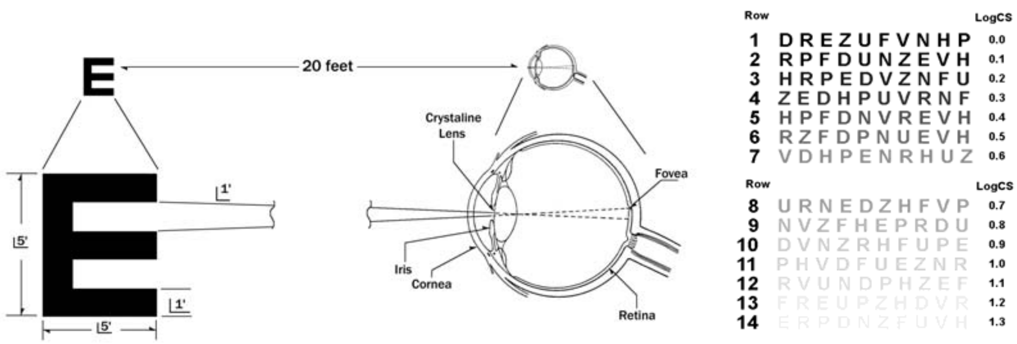


Figure 2. Acuity. The general structure of the eye is diagramed along with a basic small letter low contrast acuity chart. The letter features would be discernable at 20 feet (ft) by an individual with 20/20 Snellen acuity. Notice that the chart has both high and low-contrast letters.

Chou and Wachler (2001) said that:

> "Infrared pupillometry has shown that, with all age groups combined, approximately 50% of patients have a pupil size larger than 6.0 mm…It is noteworthy that patients receiving ablations larger than the maximal pupil size can still experience halos. After surgery, a transition zone exists between the functional optical zone and the unablated cornea. This transition zone is a source of aberrations for the patient. The higher the attempted correction is, the smaller the postoperative functional optical zone. Since the functional optical zone represents the cornea with the desired optical correction, it is generally smaller than the ablation zone. Night vision disturbances can occur with a functional optical zone smaller than the pupil size-despite a relatively large ablation. Although patients receiving ablations larger than the pupil can have night vision problems, higher myopes with larger pupils are at greatest risk…Patients at high risk for developing pupil/ablation-related night vision disturbances may be better candidates for undergoing laser vision correction with a large ablation diameter. As of press-time, 6.0 mm to 6.5 mm ablation diameters dominate treatments for simple myopia. However as excimer laser technology evolves, surgeons will have an increasing availability to use larger ablation diameters. Munnerlyn's formula predicts that for single zone ablations in the same attempted correction, larger ablation diameters subtract more tissue. Clinicians should be mindful that excessive tissue removal during LASIK may cause ectasia (tissue slump, as with a stretched and thinned cornea)." (p.7)

The incidence of problems is also unclear and somewhat dependent on the diameter of the ablation zone (Martinez et al., 1998).

> "Depending on the magnitude of the attempted correction and the size of the ablation zone, past PRK studies have reported 15% to 60% of patients complaining of glare, 26% to 78% complaining of halos, and 12% to 45% complaining of difficulty with night vision. As

many as one third of patients after PRK have been reported to be disappointed with their results despite good uncorrected visual acuity or even emmetropia. In some studies up to 10% of patients who underwent PRK with an ablation zone 4.00 mm in diameter considered the problem of halos severe enough to interfere with driving at night." (p.1053)

More recent papers have reported significant reductions in night vision problems with both LASIK and PRK (figure 1). This has come with an increase in ablation zone diameter and better ablation techniques. Of 690 questionnaires answered, 55.1% of patients reported an increase in daytime glare and 31.7% reported a decrease in the quality of night vision following surgery (Brunette et al., 2000). In spite of this, they reported that 96.2% said they believed having the surgery was a good choice. Bailey et al. (2003) surveyed 841 patients (returning questionnaires) and found a 117% increase in reporting starbursts for each 1 mm decrease in ablation diameter. In a report on a single patient, Chalita and Krueger (2004) performed wavefront-guided LASIK enhancement surgery after lifting the preexisting flap on a 3-year post-LASIK patient who presented with post-LASIK symptoms of glare, halo, and double vision. The retreatment outcome was complete resolution of double image and halos. This outcome coincided with a reduction in both low- and high-order aberrations. Chalita et al. (2004) found a strong correlation between wavefront measurements (aberrations) and visual symptoms such as coma, starbursts and glare.

Stanley Klein (2001) said that, "Night vision is an embarrassing topic for refractive surgery…A large percentage of post refractive surgery eyes have large pupils at night that result in disturbing halos." (p.51). This is particularly true for individuals in their 20's. The point-spread function (figure 3) reflects the point source image formation on the retina. Its features are directly related to blur/acuity, halo and scatter. The peak is most sensitive to blur/acuity, the shoulder to refractive-surgery-transition-zone halos, and the skirts to scatter (van den Berg and Boltjes, 1988; IJspeert et al., 1990; van den Berg, 1995; Klein, Hoffmann, and Hickenbotham, 2003).

It is clear that more factors than scatter affect night vision. The current techniques for measuring glare do not account for all of these factors. Their glare sources can produce pupil constriction; they only measure stray light; they only measure low contrast sensitivity, which can be enhanced with a glare source; they do not include measurements related to the shoulder of the point-spread function, therefore providing little or no information on halos. Consequently, there is often a poor correlation between measurements made with current disability glare measurement devices and visual performance. There is a need for better measurement strategies to evaluate and predict post-refractive surgery vision at night, particularly for young individuals with large pupils, when their vision is under conditions of low illumination (Sagawa, 1992; Klein, 2001; Klein, Hoffmann, and Hickenbotham, 2003; Barbur, 2004). Young soldiers fighting at night often have large pupils.

Complexity of the disability glare problem

Vision is a dynamic process. Nowhere is this more obvious than in the long history of efforts to develop a practical definition and measure of disability glare. We all know it when we

see it—haze, halos, reduced contrast, and starburst patterns that can obscure objects due to oncoming automobile headlights at night. Outwardly, it seems simple enough, but definition, quantification, and reliable prediction are very difficult. In fact, it is very complex on several levels: the anatomical/structural level (figure 2), the sensory level, the physics and environmental levels, and the cognitive and perceptual level. Disability glare is the overall consequence of a myriad of changing, interactive factors that can increase inter-subject variability and mask retinal image degradation (Chisholm et al., 2003).

All the following factors impact susceptibility to disability glare and/or its measurement (Bron, Tripathi, and Tripathi, 1997; Atchison and Smith, 2000; Korb et al., 2002; Kaufman and Alm, 2002). Readers who are not familiar with the details of intraocular scatter should read the following sections. Individuals that are familiar with the details can skip to the section on "Current strategies for measuring disability glare," page 15.

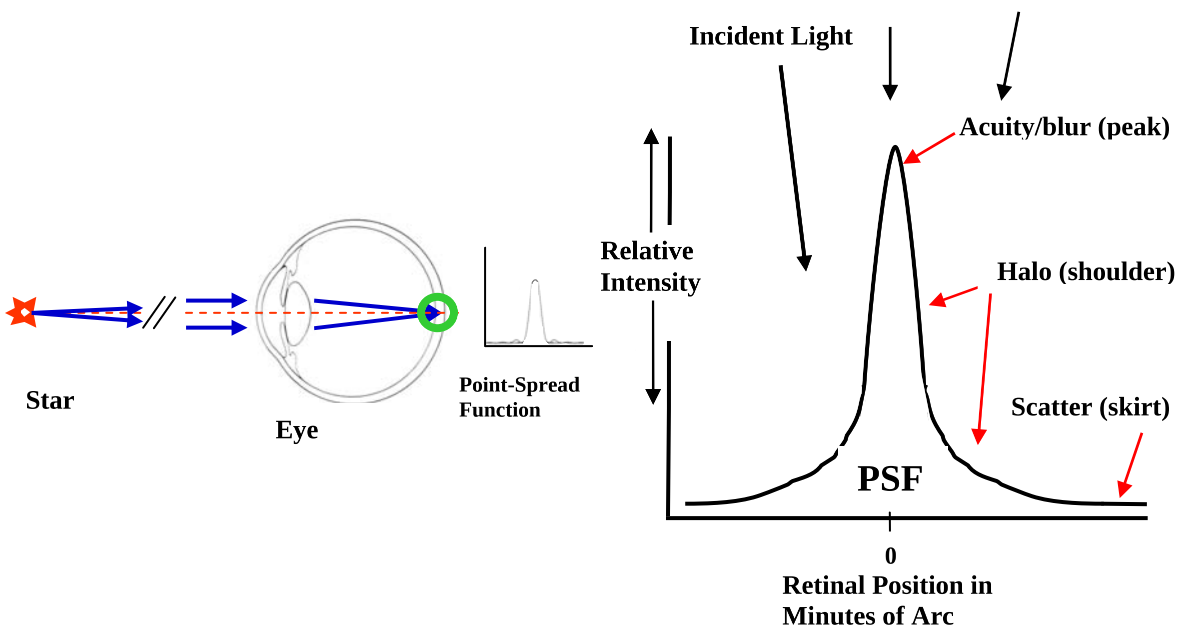


Figure 3. Point-Spread Function (PSF). A point source, like a star, does not form a point image on the retina. Diffraction, scatter, blur, and aberrations spread the light out. The function that describes the distribution of light incident on the retina from the point source is called the PSF. Defocus, aberrations, and scatter effect some regions of the PSF somewhat more obviously than others. There is a direct mathematical relationship between the PSF and the modulation transfer function of the eye.

<u>The anatomy and physiology are themselves very complicated</u>

The cornea is a multi-layered armature that forms the surface on which the tear layer, the first optical surface, forms and reforms through blinking (Bron et al., 1997). The tear layer/cornea provides more than 2/3 of the optical power of the eye (Rosenfield, 1998; Atchison and Smith, 2000). Multiple glands in the lids and around the eye socket form the complex tear

layer that has both aqueous and oily components (Bron, Tripathi, and Tripathi, 1997; Korb et al., 2002). The power of this aspherical optical surface consequently varies as a function of time. In addition, irregularities in the shape and curvature of the cornea/tear layer can create variations in focus that are very pronounced with this higher power optical surface. These variations are called astigmatism and aberrations. The corneal stroma (the central layer of the cornea), scatters about 10 percent of the visible light striking it. The stroma consists of many cross-layers of precisely spaced collagen fibers (Bron, Tripathi, and Tripathi, 1997). Their spacing is partially maintained by the tear layer, anterior cell layers called the epithelium and a single cell layer posterior to the stroma called the endothelium. They help to maintain, among other things, the correct level of stroma hydration, thereby ensuring transparency (Bron, Tripathi, and Tripathi, 1997; Korb et al., 2002; Smith, 2002).

The crystalline lens is a multi-layered optic just behind the iris/pupil (Bron, Tripathi, and Tripathi, 1997). It grows by increased layering throughout life. The crystalline lens is suspended, like a trampoline, by the zonule fibers. Sphincter muscles attached to the zonule fibers can pull on the lens to change its shape and, therefore, the eye's focus (the process of accommodation). Over the years the lens increases in size and stiffens, losing its ability to change focus or accommodate (presbyopia), as well as much of its clarity (Bron, Tripathi, and Tripathi, 1997; Ciuffreda, 1998; Rosenfield, 1998). Changes in the fibers and proteins of the crystalline lens can produce local areas that scatter light (cataract), sometimes becoming opaque (Hemenger, 1990; Swanson, 1998; Thomson, 2001). Cataracts produce primarily Mie scatter (figure 4).

Aberrations, caused by shape irregularities of the eye's optical surfaces, are partially counterbalanced by the cornea-lens optical combination (Kelly, Mihashi, and Howland, 2004; Artal et al., 2001), but this balance can be upset by both age and refractive surgery (Artal, Berrio, and Guirao, 2002; Artal et al. 2003). The power of the cornea and the accommodated lens combine with the distance between the lens and the photosensitive retina to determine whether the eye will be emmetropic (requiring no corrective intervention to focus an image).

The pigmented iris, between the cornea and the lens, is an aperture stop that reduces intraocular stray light, primarily through absorption (Keating, 1988; van den Berg et al., 1991). The pupil, the physical opening in the iris, can change diameter as a consequence of changes in illumination, convergence of the two eyes, accommodation, and emotion (Lowenstein and Loewenfeld, 1969; Bron, Tripathi, and Tripathi, 1997; Ciuffreda, 1998). In the dark, the pupils of young adults can be well over 7 mm in diameter (Schumer, 2000). On-average, pupil diameters become smaller with age. A lightly pigmented iris will absorb incoming light less and can allow some forward scatter (Voss, 2003).

Within the eye, between the cornea and the lens, is a circulating fluid called the aqueous humor. Under some conditions the aqueous humor can contain particulates that scatter light. The aqueous humor provides nutrition and oxygen for the avascular cornea. Posterior to the lens is the vitreous humor, a gel/liquid. With age, this substance begins to have localized areas of different refractive index due to pockets of localized liquefaction. These variations from

homogeneity act as local optical surfaces in the main optical pathway of the eye and can cause scatter (Bron, Trapithi, and Trapithi, 1997; Smith, 2002).

The multilayered inner surface of the eye, the retina, is a specialized extension of the brain with some 130 million or more photoreceptors of various kinds (Bron, Tripathi, and Tripathi, 1997). The photoreceptors reside in a back layer of the retina behind the retinal nerve cells. Posterior to the photoreceptors is the pigment epithelium, a pigmented layer that helps to reduce stray light in the eye, and the choroid, a highly vascular layer. Complex, laterally interacting neurons lay anterior to the photoreceptors (toward the incoming light), except in the fovea centralis, which is a tiny depression in the retina about 1.5 mm in diameter, where most of the neuron cell bodies are pushed aside. Much of our sharp vision takes place in this area (Bron, Tripathi, and Tripathi, 1997; Schwartz, 1994). The photoreceptor cells of the eye are of two general types. The more densely packed, thinner cells are called cones. They are most concentrated in the fovea and parafovea regions, rapidly diminishing in density peripheral to these areas. Cones function under brighter lighting conditions, provide our sharpest acuity, and are responsible for the first stage of neural color processing. The larger photoreceptors are called rods. They are absent in the fovea, increase in density peripheral to the fovea and then begin to decrease in density. These cells are very sensitive to light and motion, but provide much poorer acuity than cones. They do process brightness variations, but do not process color information. Photoreceptor cells change their sensitivity and range of sensitivity to light as ambient lighting conditions change--adaptation level. Initially, this process can be very rapid (Boynton, Bush, and Enoch, 1954; Bron, Tipathi, and Tipathi, 1997; Schwartz, 1994). Cones and rods interact neurally in complex ways. This is particularly important when considering lower illumination levels (Krizaj, 2000; Krizaj and Hawlina, 2002; Stabell and Stabell, 1979, 1998).

As the light environment changes the eye adjusts. Vision, as living in general, is a dynamic, changing process. When the ambient illumination increases, the eye light adapts; when illumination decreases, the eye dark adapts (Baker, 1949; 1953; Bartlett, 1966a, 1966b; Graham, 1966a, 1966b; Barlow; 1972; Schwartz, 1994). The time course for these processes is also a variable, with the most rapid changes occurring during the first few seconds of an illumination change. Other chemical, neural, and mechanical changes combine with sensory and perceptual neural processing within the brain that help to maintain a relatively stable representation of the world visually (Schwartz, 1994).

The effect of scattered light may be enhanced under conditions of low light adaptation. Intraocular stray light can cause a dark-adapted retina to light-adapt, producing a prolonged reduction in vision after the glare source has been removed. “With pathologically increased dark adaptation the effect can be stronger” (van den Berg, 1991). Steady stimuli, producing scattered light that acts more as an adapting stimulus (altering the state of adaptation), can create a paradoxical increase in contrast sensitivity as ambient light increases with low levels of illumination (Bichao, Yager, and Meng, 1995). In general, transient light stimuli are considerably more effective at producing glare and raising thresholds than are steady glare sources (Bichao et al., 1995). Under some conditions there can be a persisting visual after-image (following light stimulation), particularly in a relatively uncluttered field-of-view (Brown, 1966).

This after-image can be a result of retinal and/or central neural activity (Shinsuke, Kamitani, and Nishida, 2001).

In general we see with cone receptors above approximately 3.4 cd/m^2; these brightness levels are photopic. Between about 0.034-3.4 cd/m^2, moonlight, we see with both rods and cones; these brightness levels are mesopic. We see only with rods at brightness levels below 0.034 cd/m2; these brightness levels are scotopic. Most of us are using photopic or mesopic vision at night while driving a car.

Ultimately, one million neural fibers from each eye are sent to an area of the brain called the lateral geniculate nucleus. From this nucleus on, there is a continuing cascade of neural processing within the brain (abstracting and assembling information originating at the retina). This results in a representation of the external world that combines with memory, other senses and emotion (which can change pupil size), forming a context for behavior appropriate to our biological niche. Glare can interrupt this process.

<u>The physics of the eye's image formation is also complex</u>

The strongest refracting optical surface in the eye, the tear layer, is continuously changing and being reestablished, due to evaporation and blinking, a complicated collection of physiological processes in themselves (Bron, 1997). Transparency of the corneal stroma is maintained by light interference that depends on the separation of collagen fibers (Elliott, 1998; Atchison and Smith, 2000; Smith 2002). This separation can be affected by edema, infection, and refractive surgery. Variations in spacing can produce light scatter. Diffraction of incoming light is a function of pupil size, which changes diameter as a function of light level, emotion, and viewing distance. Below 2 mm diameter, diffraction largely defines the point-spread function (PSF) at the retina, a function describing the light distribution that results from imaging a point source object like a star (figure 3). Beyond a 2 mm diameter, focus, scatter, and optical aberrations increasingly contribute to the point-spread function (Atchison and Smith, 2000; Klein et al., 2003; Frannsen et al., 2006).

Many mathematical descriptions of a PSF assume symmetry (Atchison and Smith, 2000). The integration of the points making up an extended image also frequently assumes homogeneity (Liang and Westheimer, 1995; Westheimer and Liang, 1995a, 1995b; van den Berg, 1995). Neither of these assumptions about the image is always true due to corneal shape asymmetries, asymmetry of the pupil, and inhomogeneity of the fluids/gels within the eye, along with other factors.

Scatter adds a veil of light over an image, thereby reducing its contrast and elevating the skirts of the point-spread function (Atchison and Smith, 2000; Klein, 2001, 2003; Liang and Westheimer, 1995; Westheimer and Liang, 1995a, 1995b). Scatter can be wavelength dependent if scatter particle sizes are near the wavelengths of visible light. This is Raleigh scatter, which produces the blue sky. Scatter can also be wavelength independent if the particles producing the scatter are significantly larger than the wavelengths of visible light. This is Mie scatter, which produces the white color of clouds. Raleigh scatter produces scatter in all directions, whereas

Mie scatter tends to be more forward-directed, i. e., in the direction of the producing light, or incoming light in the case of the eye (figure 4). In the eye this is called forward scatter. Light scattered back through the pupil is called back scatter. There is also a transition in which the proportion of Raleigh and Mie scatter can vary (Hecht and Aajac, 1987; Keating, 1988; Elliott, 1998; Atchison and Smith, 2000).

The quality of an optical surface can be defined in terms of its geometrical smoothness. Zernike polynomials provide one way of systematically specifying variations in a spherical lens surface (Liang et al., 1997; Navarro et al., 1998; Applegate et al., 2001; Thibos, 2001; Thibos and Applegate et al., 2003; Thibos and Hong et al., 2002; Applegate, Hilmantel, and Thibos., 2003; Campbell, 2003). Shape variations, aberrations, result in variations in lens power. They result in relative blur and diminished contrast of images from spreading the imaged light spatially, i.e., rather than concentrating each conjugate image point of an object, each image point is spread out a little (point spread function). The main types of variations that are important for human vision are the higher order aberrations of astigmatism, sphere, coma, and trefoil (Kelly et al., 2004; Marsack et al., 2004). The last three commonly increase with refractive surgery (Martinez et al., 1998; Oshika et al., 1999; Seiler et al., 2000)

Most of the scatter from the iris and sclera is Mie scatter, although some is wavelength dependent (van den Berg et al., 1991). About 30 percent of scatter is produced by the cornea, about 30 percent by the lens, and about 30+ percent from the fundus (the inner surface of the globe). Only a very small percentage comes from the globe wall and iris, particularly with individuals having a high level of pigmentation (figure 5).

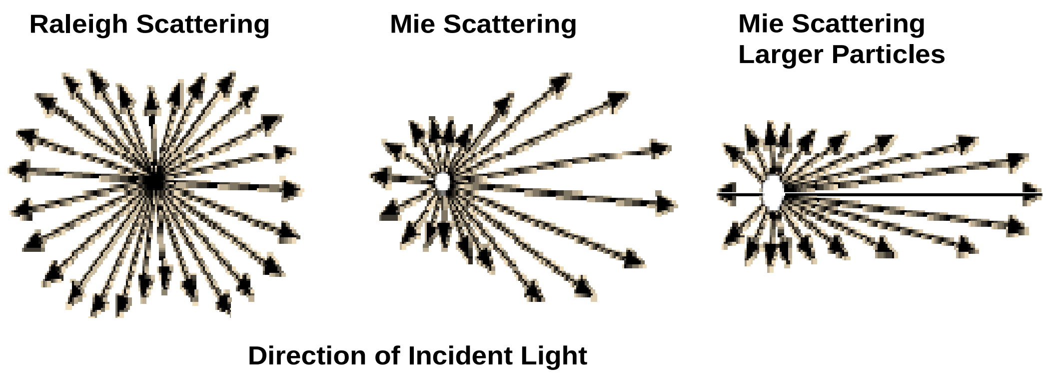


Figure 4. Scatter. Raleigh scatter, resulting from particle diameters near the wavelength of the incident light, scatters light in all directions. Mie scatter, on the other hand, tends to scatter most of the incoming light in the forward direction. The forward scatter angle narrows somewhat as the particles that create it get larger. There can be a mix of both Raleigh and Mie scatter.

There are also many factors which contribute to glare sensitivity that can exacerbate the effects of LASER refractive surgery. Any corneal edema will change the distance between collagen fibers in the stroma, increasing scatter and reducing clarity (Maurice, 1957; Jaanus, 1995; Quantock, 2000; van de Pol et al., 2001). This can occur from contact lens wear (Jones and Jones, 2000) or from a variety of diseases that affect the cornea, some acute and common like bacterial or viral keratitis (Rapuano and Heng, 2003). Dry eye, a common symptom of refractive surgery, can increase the severity of glare (Melki et al., 2003), as can trauma to the eye, a variety of ocular and systemic diseases, and allergic reactions (Bartlett and Jaanus, 1995; Kanski, 2003; Rapuano and Heng, 2003). There are many medications and drugs that can affect glare sensitivity (Blaho Connor, and Winbery, 1998). These include the obvious ones like mydriatics and cycloplegics, as well as some antiglaucoma drugs and antiallergy and decongestant drugs (Bartlett and Jaanus, 1995). In fact, the list is a long one. Both alcohol and marijuana significantly delay recovery time from glare (Adams et al., 1978).

Intraocular glare results from and combines with many environmental (extraocular) factors. The most obvious are the color, brightness, temporal characteristics, and angle (with respect to the observer's line-of-sight) of the glare source (Vos and van den Berg, 1999). But there are many other environmental factors- scratched windshields or windscreens, eyeglasses or goggles, contact lenses, type of contact lenses, fog, rain, snow and ice, time of day or sun angle, other objects like automobile chrome, flashes at night, use of night vision devices, the context in which glare occurs, and more. Each of these factors or combination of factors can influence the degree and importance of glare (Applegate, 1987, 1989; Elliott Mitchell, and Whitaker, 1991; Lewis, 1993; Pitts, 1993; Elliott, 1998; de Wit and Coppens, 2003).

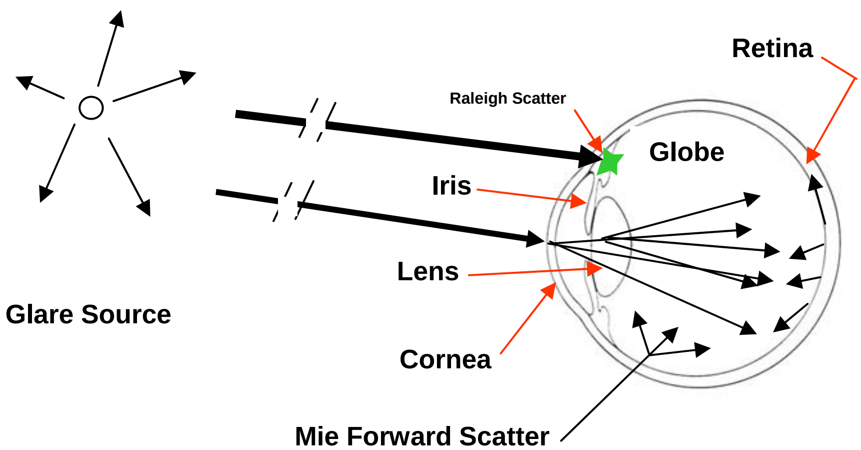


Figure 5. Scatter in the eye. The cornea, lens, and retina contribute about equally to Mie scatter. There is very little wavelength-dependent scatter, mostly for individuals with little eye pigment.

Perceptual and cognitive factors combine with sensory input to play a role in disability glare

Perception, cognition, and sensation all play a major role in disability glare (Pulling et al., 1980; Schieber, 1994a, 1994b; Anderson and Holliday, 1995; Allen et al., 2001; Green, 2004; Green and Senders, 2004;). Issues of target acquisition, recognition, and identification depend on contrast sensitivity, context, masking, clutter, and other factors, as well as sensory considerations like lighting and weather. A bright headlight may cause a reduction in pupil size, decreasing aberrations of the eye and improving acuity, but the individual may not see an unexpected object due to reduced light gathering ability of the eye and consequent reduced contrast. However, an expected object may be seen under the same conditions.

Disability glare is the combined consequence of a multitude of interacting factors, many of which are nonlinear. And disability glare can be dangerous. It can be dangerous when a night driver has only a split second to detect or identify someone or something on the side of the road. It can be dangerous when a pilot needs to detect or identify another aircraft or is engaged in critical, low-altitude maneuvers. With the advent of refractive surgery and its increasingly wider application in the military, the issue of glare with younger people has become real. There is no current gold standard for measuring disability glare and predicting problems from glare at night. At this time it is not certain that a disability glare testing strategy is even the best approach to predicting refractive surgery-caused night vision problems. It may even be that a version of wavefront analysis will prove effective and sufficient.

Current strategies for measuring disability glare

> "Over the years, many glare testers have been developed. Most of these measure either visual acuity or contrast sensitivity in the presence of a glare source. None of these has evolved into a universally accepted standard. The stray light meter provides a direct measure of intraocular stray light, instead of measuring its effect on perception. It is therefore considered the current "gold standard," but it is, as yet, suited for laboratory use only" (van Rijn et al., 2005, p.345).

The currently available methods of measuring disability glare fall into five main categories: 1) glare tests using high and/or low contrast acuity targets, 2) glare tests that try to simulate a particular glare condition, 3) glare tests that use the direct compensation method (counter-phase flicker), 4) glare tests that use increment threshold, and 5) other. These are not mutually exclusive categories, as some of the approaches combine procedures, as with the Mesotests and the Nyktotests (described below). Both of these tests use low contrast acuity targets, Landolt Cs of various contrast, but also try to simulate mesopic lighting conditions and the glare of headlights.

Glare tests using high and/or low contrast acuity targets

The greatest number of glare tests being used today measure high and/or low contrast visual acuity with and without a glare source. The most common of these are the Berkeley Glare Tester, the Brightness Acuity Tester (MARCO BAT-1000) with the Pelli-Robson Contrast Sensitivity Chart (available through any Haag-Streit authorized dealer) or Bailey-Lovie Contrast

Sensitivity Chart, the Miller-Nadler Glare Tester (MNGT), the Precision Vision back-illuminated translucent SLOAN or Rabin Small Letter Contrast Charts with an adjunct glare source, and the Vistech MCT8000.

The BAT-1000 is a hand held device often used with the Pelli-Robson Contrast Sensitivity Chart, assessed for reliability by Elliott, Sanderson, and Conkey. (1990). A battery driven light source illuminates the inside of a reflecting hemisphere that looks like half of a ping pong ball with a central hole. The patient cups the open hemisphere to the orbit of one eye and reads letters on a distant chart through the central hole. The device has 3 average bowl luminance settings--12, 100 and 400 ft. lamberts. The untested eye is occluded. The BAT 1000 and Peli-Robson chart can be purchased for under $1000.

Nakagawara et al. (1994) evaluated the BAT-1000 using a Bailey-Lovie chart, comparing it to the MCT 8000, the MNGT, and the penlight test (a clinical chair test using penlight and acuity chart). They were trying to determine the applicability of a commercial glare tester for certification of commercial pilot applicants. They found the low and medium luminance settings on the BAT-1000 produced a pinhole effect with some subjects that enhanced visual performance with respect to the no-glare condition for the low contrast chart. The high setting showed highly variable results when compared with the other settings. None-the-less, they concluded that the BAT-1000 data was relatively consistent and stable compared to the other tests, that it was the overall device/method of choice for an aeromedical certification examination of pilot applicants. Nakagawara et al. (1994) also found that the penlight test and MCT 8000 produced a paradoxical improvement in performance with glare and was relatively less sensitive to differences between glare and no-glare conditions. The MNGT did not provide no-glare acuity and the best acuity measured under the glare condition was 20/33 Snellen. Overall they concluded that there was a need for improved standards regarding commercial glare testers.

Elliott and Bullimore (1993) assessed the reliability (repeatability), discriminative ability, and validity of disability glare tests. The standard referent for validity of disability glare testers was van den Berg's Straylight Meter. They had 24 younger subjects, mean age equal 24.3 ± 3.3 years, 22 older normal individuals, mean age equal 66.0 ± 6.2 years, and 33 early cataract patients, mean age equal 70.6 ± 8.1 years. They concluded that the Regan and Berkeley tests provided similarly reliable, discriminative, and valid measures of visual assessment of cataract. The MNGT poorly detected and measured subtle changes in the ocular media, such as early cataract, because of its large step sizes at low contrast thresholds. The poor reliability and poor discriminative ability of the Vistech MCT 8000 limited its usefulness. The BAT, with medium luminance setting, used with the Pelli-Robson Contrast Sensitivity Chart was reliable and discriminative for cataracts.

Wachler et al. (1999, p.582 & p.586) pointed out that:

> "Unlike glare testing in patients with cataract it is a challenge to demonstrate glare disability in patients who have undergone refractive surgery. Because cataract is a media opacity that scatters light, reduced visual acuity and contrast sensitivity can easily be elicited by means of glare sources such as the BAT. PRK and RK (radial keratotomy) may be associated with the media opacities in the form of haze and radial scars, respectively, but optical aberrations

are more likely to adversely affect visual function because of exposure of the peripheral transition zone….Despite the inability to demonstrate decreased visual function with a glare source, certain patients continued to complain of nocturnal glare or decreased vision at night….A proper disability glare test for refractive surgery would be one that maintains natural pupil dilation to allow maximum exposure of the cornea, which may affect visual performance. Such (a) disability glare test in refractive surgery might be more accurately termed an "aberration" test, because that is its goal: to unmask spherical aberration and its effect on visual function. A proven aberration test will help clinicians diagnose and treat symptomatic patients with negative pupillary clearance (pupil size/optical zone mismatch), decentered optical zones, and other optical abnormalities. Such a test may also better define the refractive limits of different surgical procedures."

In general, these tests showed poor reliability (test-retest and across studies) and fared poorly in many validity studies (Franssen et al, 2006). All of the tests described here constrict the pupil, making the tests inappropriate for evaluating aberrations of the peripheral cornea, a major cause of halos after refractive surgery. The Straylight Meter, the comparison referent in the Elliott and Bullimore (1993) study, does a very effective job of measuring intraocualar forward light scatter, appropriate for cataract evaluation, but does not evaluate the impact of corneal aberrations resulting from refractive surgery. None of these tests will effectively evaluate the night vision of young soldiers that have had refractive surgery.

<u>Glare tests that try to simulate a particular glare condition</u>

The OCULUS Mesotest II uses Landolt Cs of two sizes and 4 contrast levels with 6 orientations each. The Landolt Cs are vacuum metallized onto glass discs and viewed against a circular background of 0.032 cd/m^2 without glare and 0.10 cd/m^2 with the glare source, simulating automobile traffic at twilight or at night. The glare source is presented at 3^o with an intensity of 0.35 lux at the pupils. Illumination for the glare source and background is provided by a regulated halogen bulb. Fixation is assisted by two projected red spots, one above and one below the acuity target. The unit is binocular, with instrument myopia "eliminated for the most part," and viewing corrected to allow the eyes to "accommodate and converge as in normal vision." Contrast levels, glare condition, and orientation are presented in a partially randomized sequence. The unit can be computer-controlled, allowing modifications of the sequence and targets used (OCULUS Mesotest II Instruction Manual; Baldwin, 2001). The basic model sells for a little over $7000 in the United States. A very similar device, the Rodenstock Nyktotest 300 is not readily available in the United States. Consequently it is not widely used here. Both the Mesotest and the Nyktotest are more frequently used in Europe and have been considered for drivers license testing. A recent comparison and evaluation of the Mesotest II and the Nyktometer was conducted by van Rijn and Nischler et al. (2005). They compared results from 40 young (20-40 years old), 37 elderly (50 years old and older), and 35 cataractous patients. Age, health, refraction and acuity were controlled. Their results indicate some of the major reasons that they are not employed in driver vision evaluation testing. They found that the small difference between normal and impaired responses was too small given the standard deviation of measurements. In addition, young, healthy eyes gave false positives, and some cataract patients passed (possibly not a false negative). Van Rijn and Wilhelm et al. (2005) evaluated 93 subjects

50 years old and over using the Mesotest II and Nyktotest 300 as a screening test for drivers and related their results to a vision questionnaire. Ninety-one of these subjects were active drivers. Perceived driving disability (PDD) was calculated from the questionnaire, which was shown to be significantly dependent on visual acuity, contrast sensitivity and useful field of view, as well as the Mesotest II with and without glare and the Nyktotest 300 with glare. It should be noted, however, that the tests were restricted in their usefulness to night and bad weather conditions. Both glare tests had a high failure rate of about 40% with glare.

Glare tests that use the direct compensation method (counter-phase flicker)

This procedure makes the assumption that disability glare is the same as, or a high correlate of forward intraocular scatter. It is usually assumed that the type of scatter is Mie scatter, i.e., having no wavelength-dependent component, although lights with a specific hue can be incorporated into the test (van den Berg et al., 1991; De Wit et al., 2005a, 2005b). There are two types of direct compensation stray light measurement devices currently being used: first, variations of van den Berg's Straylight Meter (Elliott and Bullimore, 1993; van den Berg, 1995; Smith, 2002; Vos, 2003; van Rijn, Nischler et al., 2005), and second, a computerized stray light meter developed most fully along with the European Glare Study (Harrison et al., 1995; De Wit, 2002; De Wit et al., 2005a, 2005b; van Rijn, 2002; van den Berg and van Rijn, 2005; van Rijn, Nischler et al., 2005; van Rijn, Wilhelm et al., 2005). The Straylight Meter has a long and well-documented history of effective and reliable measurement of intraocular stray light in a laboratory setting (Elliott and Bullimore, 1993; De Wit et al., 2005a, 2005b; van Rijn, Nischler et al., 2005; van Rijn, Wilhelm et al., 2005; Franssen et al., 2006) and there is growing evidence that the computerized versions do as well as the non-computerized versions (Harrison et al., 1995; de Wit et al., 2005a, 2005b; van Rijn,Nischler et al., 2005; van Rijn, Wilhelm et al., 2005). The major problems with the direct compensation method that prevented it from general clinical use are: 1) difficulty of naïve subjects to reliably judge weak flicker, 2) a contraintuitive task of determining a no-flicker end point, 3) measurement accuracy dependent on the observer's adjustment strategy, 4) no control over a subject's measurement reliability, 5) a subject's ability to influence measurement outcome (Baldwin, 2001; Franssen et al, 2006).

A version of the Straylight Meter, the C-Quant™ was first produced commercially in 2005 by OCULUS (de Wit, 2005a, Franssen et al, 2006). The end-point of the task for this device has been redefined from a direct compensation method (minimum flicker state) to a compensation comparison method ("which of two sides flickers most"). This is purported to make stray light measurements more reliable and applicable to the clinical setting (de Wit, 2005a; Franssen et al, 2006).

There is little doubt that the Straylight Meter is effective for measuring intraocular forward scatter in the eye and has been widely tested for evaluating cataracts and age-related intraocular scatter. It is sensitive to any of the scatter-related components of disability glare (Franssen et al, 2006). However, as far as the authors here are aware, there is no strategy for modifying the Straylight Meter to be sensitive to the shoulder of the point-spread function ($< 0.5^{o}$), thereby making it sensitive to aberrations of the cornea. This, of course, does not preclude its use with

other measurement devices to get a more comprehensive indication of an individual's night vision performance.

Glare tests that use increment threshold

Increment threshold is a psychophysical procedure where the luminance of a superimposed test spot is compared to that of its surround by an observer. The surround is often, but not necessarily, a homogeneously filled circle with a spot presented at its center. If the spot intensity is just detectable against its surround the increment of luminance difference (ΔI) between the spot and its surround is said to be one just-noticeable-difference (jnd). A function describing the relationship between an increasing, suprathreshold surround luminance and the corresponding, increasing jnd or ΔI is called Weber's Law. It states that the ratio of ΔI to the log background luminance is constant. This relationship generally holds over a 4-log unit luminance change beyond the dim light level where the quantal fluctuations of the light source have a major impact on determination of threshold.

As far as the authors are aware, there are no current commercial disability glare testers that use a variation of this psychophysical procedure; most use some sort of contrast target. However, it was used frequently in early pioneering research (Voss.2003; Franssen, 2006) and has been used in recent research (Applegate et al, 1987; Wooten and Geri, 1987; Harrison et al 1995; Westheimer and Liang, 1995b). The main reason that it has not been used in commercial glare testers is the difficulty naïve observers have determining a reliable threshold. In research it is possible to use well-trained (sophisticated) observers that are able to spend a great deal of time on the task. It remains to be seen whether it can be eventually adapted for clinical use.

Glare tests that use a variation of the increment threshold for glare/scatter evaluation go back to the work of Cobb, who developed the concept of equivalent veil. Applegate et al. (1987) used a two-channel optical system to create a 5^o square background with a point glare source set at 1^o-2^o from a central 0.5^o variable test spot. They measured "large glare effects [for radial keratotomy patients]…at low background luminance levels." Harrison et al. (1995) compared one-month post PRK forward scatter measurements from the Stray Light Meter, the Computerized Stray Light Meter, and an increment threshold-glare paradigm using descending method of limits. They found that the differences among techniques were small and clustered around zero. They also found data obtained from the increment threshold procedure had the least variability, although none of the procedures showed a high standard deviation. Wooten and Geri (1987) used another variation of the increment threshold procedure to determine if intraocular light scatter was wavelength dependent. They concluded it was not.

Westheimer and Liang (1995b) also used an increment threshold procedure to get an estimate of ocular scatter. The increment threshold psychophysical procedure that they used is diagramed in figure 6. A fixed uniformly filled white circle or annulus with a 20^o outer radius was used that ranged from 6-40 cd.m^2. In order to obtain proportional data within 7 min of arc of the PSF peak Westheimer and Liang (1995b) patched results obtained from their double-pass measurements (Liang and Westheimer, 1995; Westheimer and Liang, 1995a) onto the PSF derived from the increment threshold method. A schematic of the double-pass procedure is

shown in figure 7. No effort was made to evaluate effects of aberrations or blur on the PSF. It was assumed that the physical component causing the point-spread function was primarily Mie scatter. They argued that their increment threshold procedure reflected characteristics of at least the skirts of the point spread function (beyond 30 min arc), a region particularly sensitive to stray light. Whether this procedure can be used effectively with mesopic light levels is not known.

It is possible that increment threshold data obtained using a central spot and annulus can be contaminated by lateral interaction of retinal neurons. Data obtained by Westheimer (1967) showed that this interaction does not extend beyond ~ 12 min for fovea cone vision with their increment threshold experimental setup. The inner diameter of the annulus used by Westheimer and Liang (1995b) is well beyond this, 30 min or greater, eliminating the possibility of lateral neural interaction. However, a procedure developed by Klein et al. (2003) used an arrow tip very close to this limiting distance (discussed below).

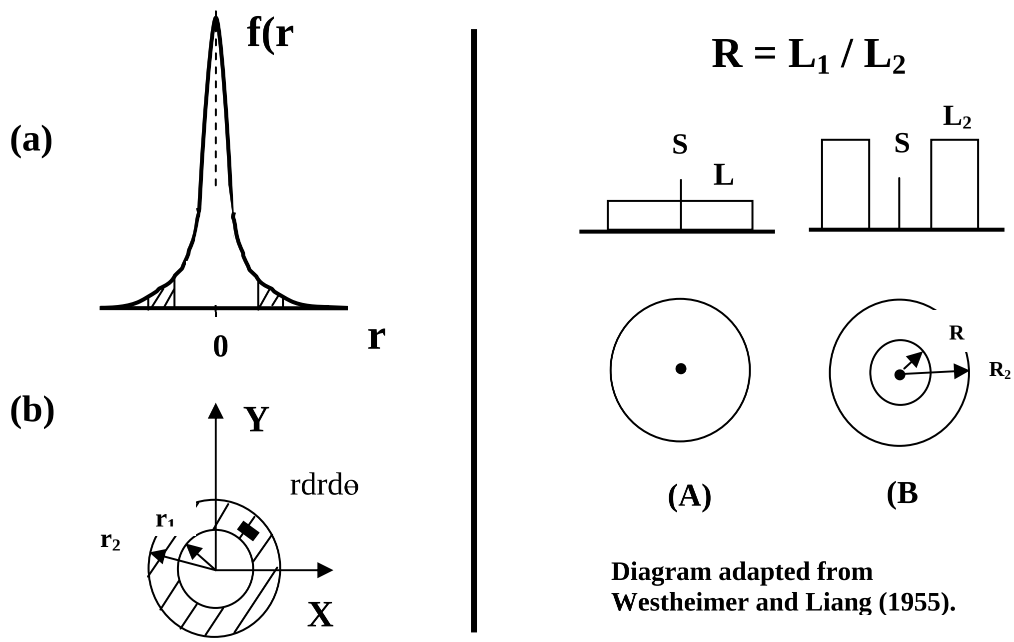


Figure 6. Westheimer and Liang increment threshold procedure. This illustrates the principle underlying the increment threshold procedure used by Westheimer and Liang (1995b) to estimate intraocular forward light scatter from a point source (left) and their experimental procedure (right). Left--A normalized circularly symmetrical, 3-dimensional PSF f(r) at (a) is the conjugate retinal image related to a luminous point object. The same coordinate system is used for the conjugate object and retinal image planes. That the conjugate object and retinal image are related by a constant is ignored. Symmetry of the PSF is assumed. The total light flux (E) from the point object incident on the annular zone with inner radius r_1 and outer radius r_2 is given by

$$\mathbf{E(r_1, r_2) = \int_{r_1}^{r_2} f(r)2\pi r dr.} \qquad \mathbf{(3)}$$

The total light contributed by each element of area $rdrd\theta$ on a homogeneous annulus target of unit intensity bounded by inner radius r_1 and outer radius r_2, that falls at the image center is $f(r)rdrd\theta$. Consequently, the total light intensity (I) contributed to the center of the annulus by this target relative to the intensity at each point in the annular zone is

$$\mathbf{I(r_1, r_2) = \int_{r_1}^{r_2} f(r)2\pi rdr.} \qquad \mathbf{(4)}$$

"Because the two integrals are identical, it follows that the intensity at the center of the image of an annular object pattern bears the same proportion to the annulus intensity as the total flux contained in the same annular zone of the image of a point object does to the total volume under the whole PSF." Right--a single white spot was flashed every 100 ms at the center of a screen with either a white homogeneously filled circle or annulus. The luminance of all three was controlled by neutral density (ND) filters. The threshold of the spot was obtained by the observer increasing or decreasing the luminance of the disc or annulus. A comparison of the disc and annulus luminance at spot threshold provided a proportional estimate of the forward scatter.

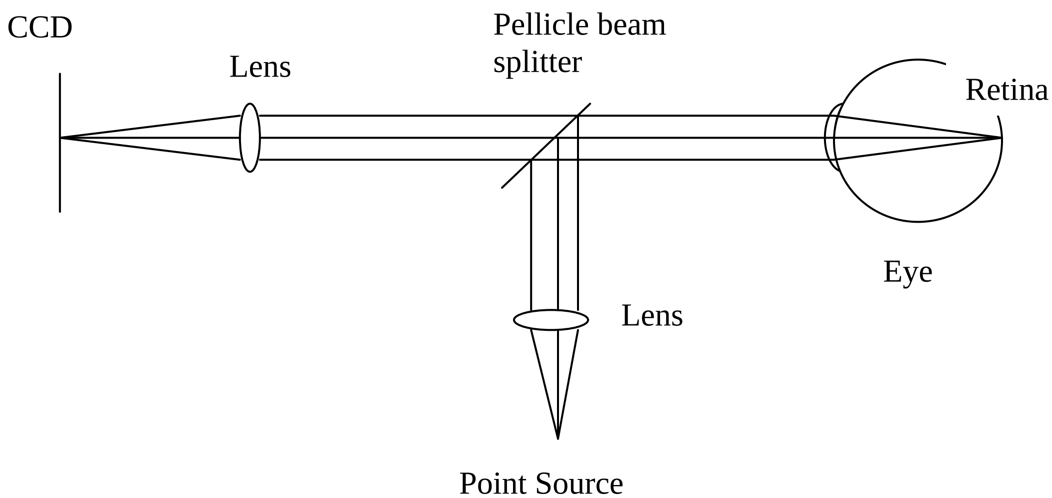


Figure 7. Schematic of double pass procedure adapted from Liang and Westheimer (1995ba) and Westheimer and Liang (1995a and 1995b). Monochromatic light from a point source is collimated and reflected via a pellicle beam splitter to the eye, where it is focused on the retina. Light reflected from the retina is re-collimated by the optics of the eye, passed through the pellicle, and focused by a lens onto the CCD camera plane.

<u>Other glare testing strategies</u>

Discussion of three relatively new procedures for determining disability glare and halos follows. The first of these is a direct method developed to assess halos following LASIK surgery (Gutierrez et al., 2003). They presented a flat panel with 12 myridians, each with 10 diodes

spaced away from a center diode. The patient was dark adapted and instructed to view the center of the panel. The number of lighted diodes not obscured was compared to the total number, providing a percentage index of the visual deficit. A comparison of pre to post surgery results showed a significant decrease in the index percentage. There are a number of ways this device might be used, including varying the spot luminances to a level of detection. It is clear that this device can provide an inexpensive way of evaluating significant halos. It has not been shown to be an effective way of evaluating intraocular stray light or evaluating disability glare. However, in conjunction with the Straylight Meter, it may predict night vision deficits. Further development and evaluation of the technique may be justified, including ways to reduce testing time, ways to eliminate the need for a dark environment, and ways to reduce learning effects.

A second approach for evaluating disability glare has been proposed by Klein, Hoffmann, and Hickenbotham (2003). Their technique uses an increment threshold strategy, but with a central arrow replacing a central spot. The tip of the arrow is positioned so that the tip is within 15 min of an inner annulus edge. Luminance levels are maintained in the low photopic-mesopic range. The arrow can be rotated and flashed within a 1^o inner diameter annulus or 1^o pedestal (disc). An observer is required to elevate the annulus or pedestal luminance while indicating the direction the arrow is pointing. The threshold or end-point occurs when the observer can no longer determine direction pointed.

A similar procedure can be used with a Landolt C (Berman et al., 1993, 1994). The strengths and weaknesses of this type of disability glare testing procedure need to be further evaluated, particularly at mesopic luminance levels.

The third approach is more complicated than the first two. Chisholm et al. (2003) adapted their technique to the visual requirements of pilots, thus evaluating visual performance over a 5^o visual field. Following 15 minutes of dark adaptation they tested the threshold size for determining the orientation of a Landolt C randomly presented at 0^o, $\pm 1.25^o$ and $\pm 2.5^o$ (horizontal) from the line-of-sight. A fixation orientation target surrounded a dark fixation point. Two background illuminations were tested, 0.05 cd/m^2 (mesopic) and 12 cd/m^2 (photopic). Contrast levels for the Landolt Cs were 48% and 24% respectively. The contrast acuity assessment (CAA) was a plot of the Landolt C presentation angle against threshold size. Interpretation was based on normative data or shape of the plot. They also developed a rapid CAA screening test, but did not say whether 15 minutes of dark adaptation was required. Further development is needed to reduce testing time and ensure evaluation validity for this procedure.

<u>USAARL pilot efforts to develop a low-luminance disability glare tester</u>

Low luminance, low contrast Landolt C orientation study

During 2000-2001 Corina van de Pol (2001) and Victor Klymenko (2001) began pilot development of a low-luminance glare measurement procedure. The objective was to develop a pupil-sparing measure of disability glare that was appropriate for post refractive surgery patients and that predicted night vision performance. C-language software, with Hewlett Packard graphics commands, was developed at USAARL for presentation of Landolt Cs of

programmable size, contrast, orientation, and duration/duty cycle. These targets were seen against backgrounds of programmable shape, size, luminance, and duration/duty cycle. Most glare sources were generated on the monitor, although some external glare sources were tried. A transformed, interleaved up/down psychophysical method was developed to determine Landolt C orientation threshold under various non-glare and glare conditions. Feedback on response accuracy was given to observers every x^{th} trial to maintain subject motivation. No procedures were ever developed to monitor or control pupil size. However, this was a long-term goal.

Stimuli were presented in a darkened room using a non-interlaced, 60 Hz, non-glare, vertical scan, RGB, 1280 H X 1024 V Cathode Ray Tube (CRT) monitor. Although most of this early pilot data no longer exists, some representative fragments remain, two of which are shown in table 1. In two instances, successive 9.3' Landolt Cs were successively presented at monitor center, each presentation (trial) with one of four randomly chosen orientations. Stimulus presentation automatically followed the observer's selection of a response orientation. Two non-glare background runs (constant dark background) were followed by two glare annulus runs (annulus luminance fixed). Each run used a six-turn stair step psychophysical procedure. The turning points were automatically determined when the observer selected two successive correct orientations during an ascending target luminance or two successive incorrect orientations during a descending target luminance series. Observers viewed the monitor at 2.5 meters. The inside diameter of the glare annulus used for this data sample was 31.6'; the outside diameter was ~371'.

There were three major lessons-learned from this early disability glare pilot work. First, the staircase psychophysical procedure needs to be much shorter to be workable in a clinical setting with naïve observers. Second, monitoring and controlling pupil size is essential, as is tight control of target and ambient illumination. This is particularly true if the visual effects of a large natural pupil, as occurs at night, are to be tested. Third, data variability and reliability with threshold-type tasks can be a serious problem when using naïve observers. Techniques that reduce variance need to be explored.

Table 1.
Two early disability glare measurement pilot runs.

OBSERVER = J1
Glare annulus 250 luminance steps; Target start 84 luminance steps;
4, 4 step sizes (2), 40 luminance steps below anticipated end luminance;
Target practices 84 luminance steps and 92 luminance steps; 6 turnpoints; Step size reduced after 2 turnpoints;
Trial do-overs = 1/10trials
Stimulus durations for runs 1 (practice, no glare), 2 (experimental, no glare), 3 (practice, glare), 4 (experimental, glare) = 120 sec, 30 sec, 30 sec, 15 sec
Stimulus on-time for runs 1, 2, 3, 4 = 60 sec, 15 sec, 15 sec, 7 sec
Temporal frequency = 0

Total trials = 202
Non-glare practice (U = 56; non-glare experimental = 60 ; glare practice = 47 ; glare experimental = 39

	No Glare				Glare			
Turning Points	Up1	Down1	Up2	Down2	Up3	Down3	Up4	Down4
0	----	----	----	----	----	----	----	----
1	44	32	56	44	80	52	88	72
2	36	32	48	36	60	44	80	76
3	36	32	40	32	48	44	84	80
4	36	32	36	28	56	52	84	80
5	44	36	32	28	56	52	84	68
6	40	36	40	36	56	52	76	72

Means of Last 4 (after step size change)
Sequence 2 average Up = 37.0 average Down = 31.0 average MID = 34.0
Sequence 4 average Up = 82.0 average Down = 75.0 average MID = 78.5

OBSERVER = R1
Glare annulus 128 luminance steps; Target start 28 luminance steps;
4, 4 step sizes (2), 20 luminance steps below anticipated end luminance;
Target practices 44 luminance steps and 32 luminance steps; 6 turnpoints; Step size reduced after 2 turnpoints;
Trial do-overs = 1/10trials
Stimulus durations for runs 1 (practice, no glare), 2 (experimental, no glare), 3 (practice, glare), 4 (experimental, glare) = 120 sec, 30 sec, 30 sec, 15 sec
Stimulus on time for runs 1, 2, 3, 4 = 60 sec, 15 sec, 15 sec, 7 sec
Temporal frequency = 0

Total trials = 136
Non-glare practice = 38; non-glare experimental = 33 ; glare practice = 33 ; glare experimental = 32

	No Glare				Glare			
Turning Points	Up1	Down1	Up2	Down2	Up3	Down3	Up4	Down4
0	----	----	----	----	----	----	----	----
1	32	28	32	24	44	36	36	32
2	32	24	28	24	40	36	36	32
3	32	20	36	20	44	40	40	36
4	32	28	24	20	44	40	40	36
5	32	24	28	24	44	36	48	32
6	28	20	32	28	40	36	36	32

Means of Last 4 (after step size change)
Sequence 2 average Up = 30.0 average Down = 23.0 average MID = 26.5
Sequence 4 average Up = 41.0 average Down = 34.0 average MID = 37.5

Between 2000 and 2004, two pilot disability glare measurement procedures were tried by Corina van de Pol on the Accession and Rated Aviator refractive surgery studies at USAARL (Bissette, 2006; van de Pol, 2000, 2001, 2004; van de Pol, Bower, and Rabin, 2001; van de Pol, Soya, and Hwang, 2001; van de Pol et al., 2007). The first used was the Berkeley Glare Tester. In addition to unreliable performance, this test often resulted in paradoxical low contrast acuity improvement with the glare source on. This probably resulted from pupil constriction during the glare phase of the test, reducing the impact of optical aberrations. Retro-illuminated Precision Vision 5 percent low contrast charts were later used with an adjunct glare source, two side-mounted halogen sources. Paradoxical low contrast acuity improvement was also a problem with this setup. The lesson learned is clear- pupil size control during disability glare measurement is essential.

In-house Laboratory Independent Research (ILIR) Study to develop a low-luminance disability testing procedure

During the Accession and Rated Aviator studies (Bissette, 2006; van de Pol et al., 2004, 2007) it became increasingly evident that there was need for a pupil-sparing disability glare measurement procedure that was predictive of the night vision problems that many post refractive surgery patients were experiencing. A novel approach was proposed in 2003 that resulted in an In-house Laboratory Independent Research (ILIR) study, More Reliable Glare Testing (Kalich, 2004). The initial plan for this project was to determine whether the separation of two just-discernable computer monitor-generated spots could be used as an indicator of degree of disability glare. This idea was derived from the fact that blur/aberrations broaden the PSF and reduce its amplitude, requiring increased point object separation for an observer to just discern two points. In addition, a scatter caused veiling luminance would combine with the effects of blur and aberrations to require the peaks of the PSF to be further separated for two spots to be discernable. If sensitive and reliable enough, this measure could provide the basis for a single indicator of disability glare resulting from both increased post refractive surgery aberrations and intraocular light scatter, something that heretofore had not been accomplished (figure 8).

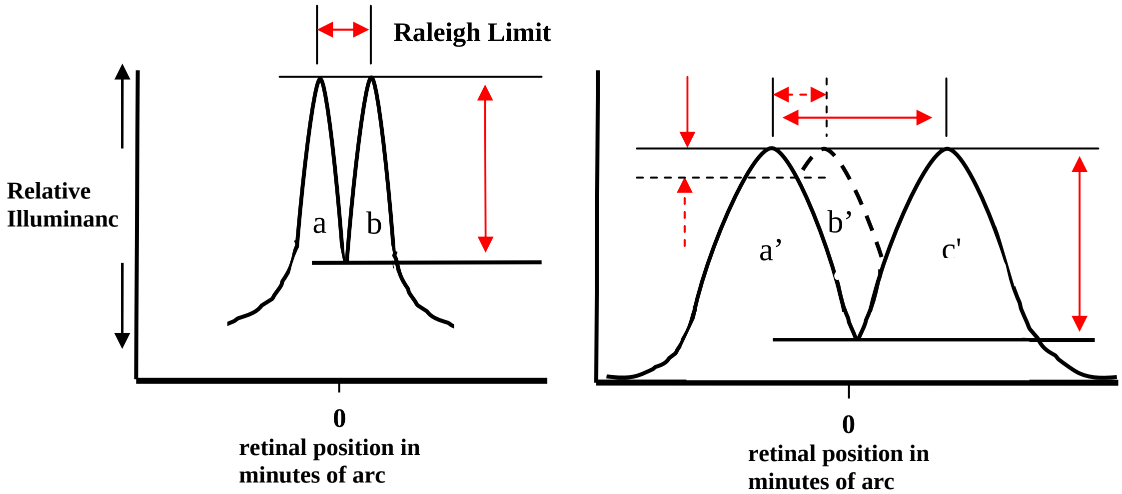


Figure 8. Two PSF discrimination. Shown are four exaggerated PSFs, a, b and a’, b’. Each image pair results from the same two point source objects. The a, b PSFs represent images resulting from a relatively aberration-free lens and scatter-free medium. They are shown to be at the Raleigh image resolution limit, meaning that an observer would be just able to distinguish that there are two point source objects. The a’, b’ PSFs on the reader’s right represent PSFs from a lens with large-magnitude aberrations or blur. In this case the observer would see only one combined spot, from a’,b’, unable to determine that there are two separate point sources. The PSF separation would have to be increased to a’, c’ for the observer to see that there are two.

The idea was tested in 2004 for computer-monitor generated spots, thus allowing a variety of conditions to be evaluated. Low-luminance white and colored spots (1 pixel minimum) were tried with observers using fogging filters, aberration waveplates (table 2), and blur lenses in combination with an artificial pupil. A pupil-sparing annulus was used as a glare source. The spots and annuli were generated on a geometry-corrected 1600 by 1200 pixel EIZO FlexScan F980 CRT monitor with an average measured pixel diameter of 0.245 mm. The monitor brightness and log luminance step profile is provided in figures 9 and 10.

Software and adjustable viewing distance provided a wide range of spot/annulus/pedestal sizes and viewing angles, colors, and luminance levels. Spots were presented at the monitor screen center. A blank monitor screen and one with a black-line-cross eye guide with an 18 pixel diameter or larger annulus center were tried. The thin black lines were just visible when a minimal background or annulus luminance was introduced and clearly visible when an annulus or pedestal glare source was employed (figure 11). A series of three gloss black nested frustums were employed in front of the monitor screen to shift unwanted screen light/scatter away from the observer (figure 12).

Eye position was monitored using a SONY Cyber-shot DSC-F717 Digital camera viewing a mirror in front of the left eye. Corrective lenses, neutral density filters, fog and light scattering filters, blur lenses and aberration waveplates were mountable in front of the right eye, centered

along the axis of the nested frustums (figures 13 and14). The left pupil was monitored by one of the experimenters using the SONY camera (figures 14 and 15). A near infrared illuminator was used for some eyes, although it generally proved unnecessary. Transmission of filters and scatter characteristics were measured using expanded LASER beams. The apparatus used to measure scatter is shown in figure 16. A scatter profile of one of the fog filters is provided in figures 17.

During extensive pilot studies, steady state, blinking or alternating computer-generated spots were presented to observers. Spot brightness was manipulated using 255 monitor brightness steps or 72 ~ equal log luminance steps (figures 9 and 10; appendix A). Two-spot separation was computer manipulated or independently manipulated by the observer in 1-pixel increments. The method of limits and a number of staircase psychophysical procedures were tried to determine threshold end points: 1) the separation at which two in-phase blinking spots with the same duty cycle were just resolved by an observer; 2) the separation at which two counter-phase spots with the same duty cycle were just seen to alternate; 3) the spot separation required to just discern two continuously-presented spots instead of one; and 4) the glare annulus luminance at which two continuously-presented or continuously-alternating spots of constant luminance and separation could just be discerned. Spot target presentations of varying color, luminance, size, and separation were tried with and without annulus glare sources of varying luminance, color and inner and outer diameters. A minimum spot angular subtense of 50 arc seconds could be achieved. Humans can determine that spots somewhat smaller than 20 seconds of arc have size. This is made clear by looking at planets in the night sky and noticing that their twinkle behavior differs from that of pinpoint stars.

The alternating spot data had the lowest standard deviation of the 2-spot pilot trials. Example data from a single researcher subject is provided in appendix B; alternation frequency and duty cycle were evaluated during these trials. However, no reliable glare-related difference between glare and no-glare end-points associated with aberrations or low blur levels was achieved during the 2-spot pilot testing, even when experimenters served as observers.

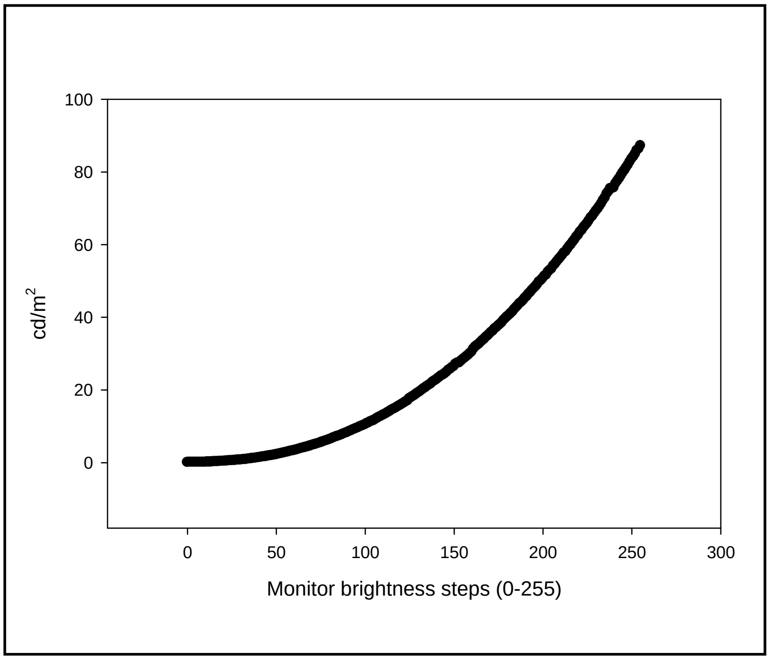


Figure 9. Monitor 256 brightness-step luminance profile.

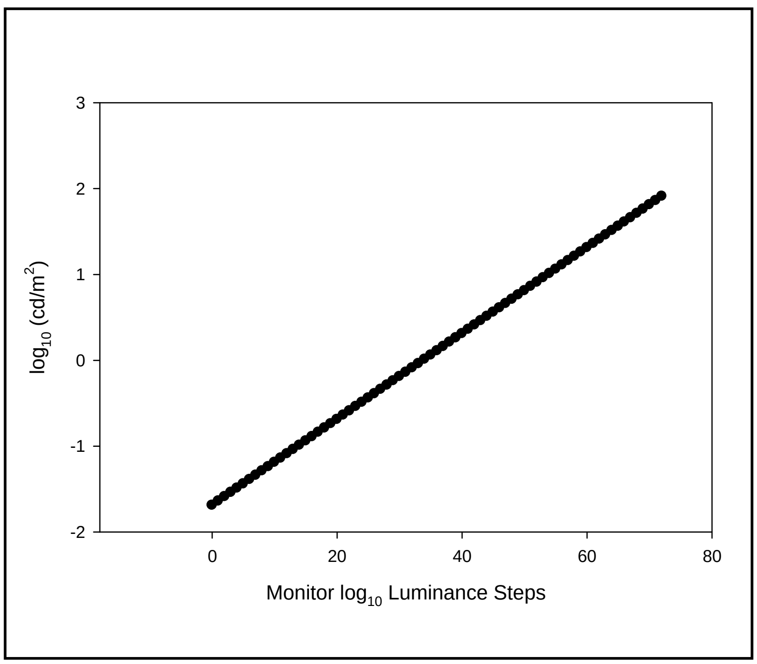


Figure 10. Monitor log luminance-step profile.

Table 2
Filters used in the ILIR disability glare study.

I. Neutral Density Filters:

1. Hoya HOND8MC27 multi-coated 8X [ND 0.86 (measured)] neutral density glass filter
2. Canon FS-H37U [ND 0.93 (measured)] neutral density glass filter
3. Tiffen 37DFK3 [ND 0.61 (measured)] neutral density glass filter

II. Fog Filters

1. Tiffen Double Fog 3 Filter (yellow)
2. Tiffen Fog 5 Filter (black)
3. Tiffen Double Fog 5 Filter (pink)

III. Aberration Waveplates Calculated for a 6 mm Aperature*

1. Ophthonix 38.3 mm dia./2.4 mm thickness Plano Waveplate
2. Ophthonix 38.2 mm dia./2.4 mm thickness Z(4.0) = +0.31 µm rms (sphere) Waveplate
3. Ophthonix 38.2 mm dia./2.4 mm thickness Z(4.0) = -0.36 µm rms (sphere) Waveplate
4. Ophthonix 38.2 mm dia./2.3 mm thickness Z(4.0) = +0.6 µm rms (sphere) Waveplate
5. Ophthonix 38.2 mm dia./2.4 mm thickness Z(4,0) = -0.67 µm rms (sphere) Waveplate
6. Ophthonix 38.2 mm dia./2.8 mm thickness Z(3,-1) = -0.4 µm rms (coma) Waveplate
7. Ophthonix 38.2 mm dia./2.9 mm thickness Z(3,-1) = -0.65 µm rms (coma) Waveplate
8. Ophthonix 38.3 mm dia./2.9 mm thickness Z(3,-1) = -0.99 µm rms (coma) Waveplate
9. Ophthonix 38.2 mm dia./2.9 mm thickness Z(3,-3) = -0.34 µm rms (trefoil) Waveplate
10. Ophthonix 38.2 mm dia./2.9 mm thickness Z(3,-3) = -0.58 µm rms (trefoil) Waveplate
11. Ophthonix 38.2 mm dia./2.9 mm thickness Z(3,-3) = -0.96 µm rms (trefoil) Waveplate

* Specially crafted to USAARL specifications and provided on loan from Ophthonix for this Glare study.

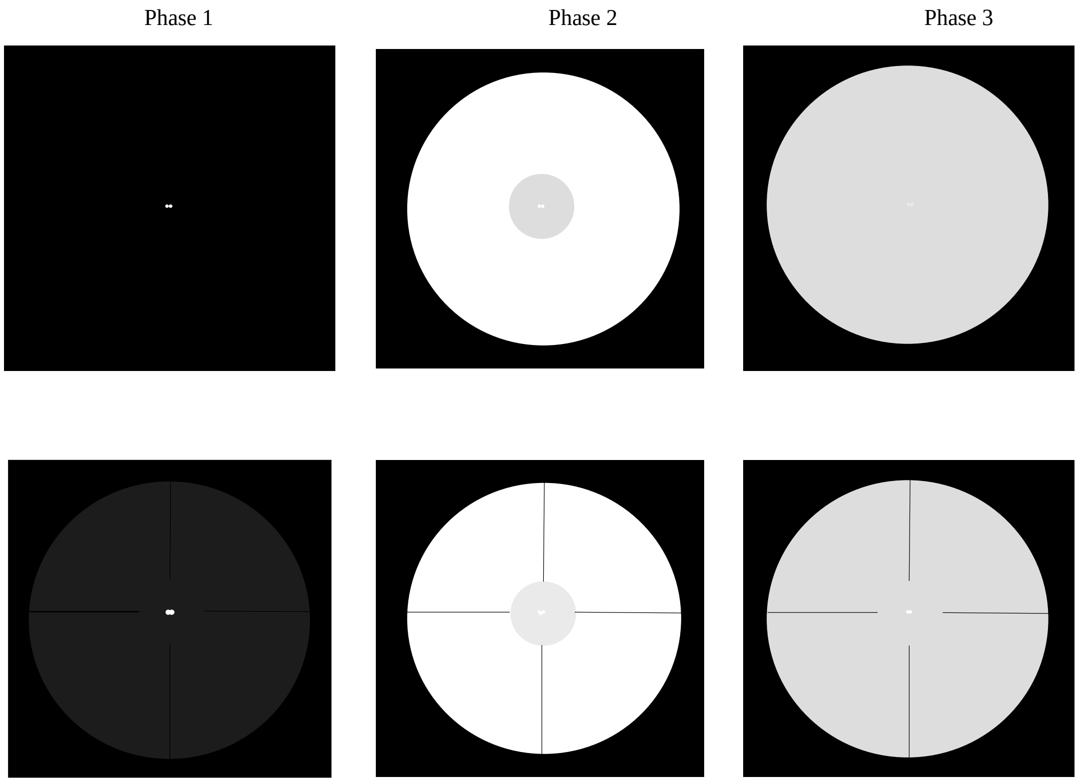


Figure 11. Three phases of spot presentation. Three screens with and without a black line cross eye guide are shown. In phase 1, the two spot luminance could be incremented until the two spots are just discerned or the separation of two constant-luminance spots could be adjusted until the two spots were just detectable. In phases 2 and 3, the annulus or pedestal luminanace could be incremented until two spots, luminance or separation determined from phase 1, could no longer be discerned. The annulus and the pedestal luminance could also be preset and the spot separation manipulated by the observer until two constant-luminance spots could no longer be detected. A light adaptation screen was presented just prior to presentation of each phase.

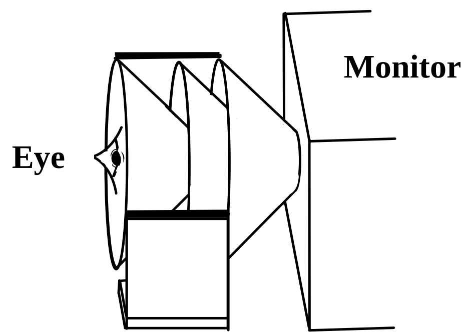


Figure 12. Nested frustums. Nested frustums direct stray light from the monitor away from the port and observer. The three frustums were supported by angle braces that also reflect scattered light away from the monitor and the viewing port.

Figure 13. Subject position in glare tester. The subject views the monitor screen spot and glare annulus along the axis of three nested frustums. Subject responses are made using a joystick.

Figure 14. View of the glare tester from subject's perspective. Looking into the frustum nearest the subject the adjustable lens holder for the right eye can be seen along with the mirror-tube assembly allowing a continuous view of the left eye. The camera viewing the eye can be seen to the left. Corrective lenses, blur lenses, neutral density filters, fog and scatter plates, waveplates and artificial pupils were positioned in front of the right eye.

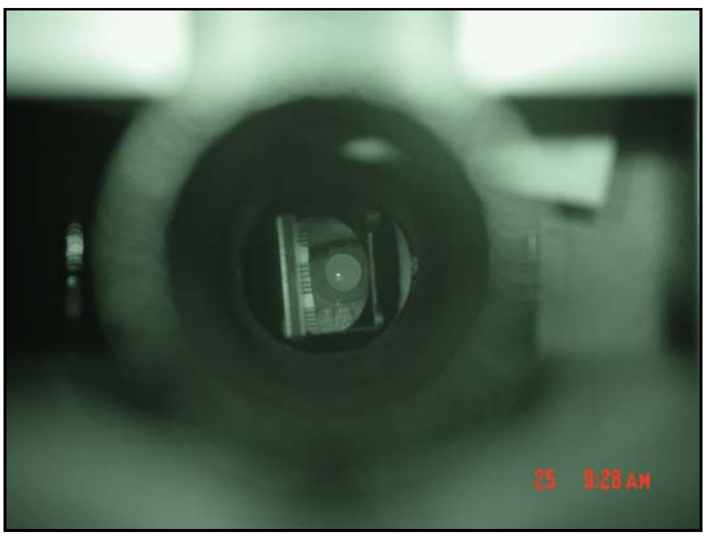


Figure 15. Camera view of a subject's left eye. In general, the pupil diameter of the left and right eyes are about the same diameter. A millimeter scale provides an estimate of the pupil size by the experimenter. A digital photograph can be taken at any time, allowing a very accurate determination of pupil size. In addition to pupil-monitoring, the position of the left eye provided a good indication of the subject's correct right eye position.

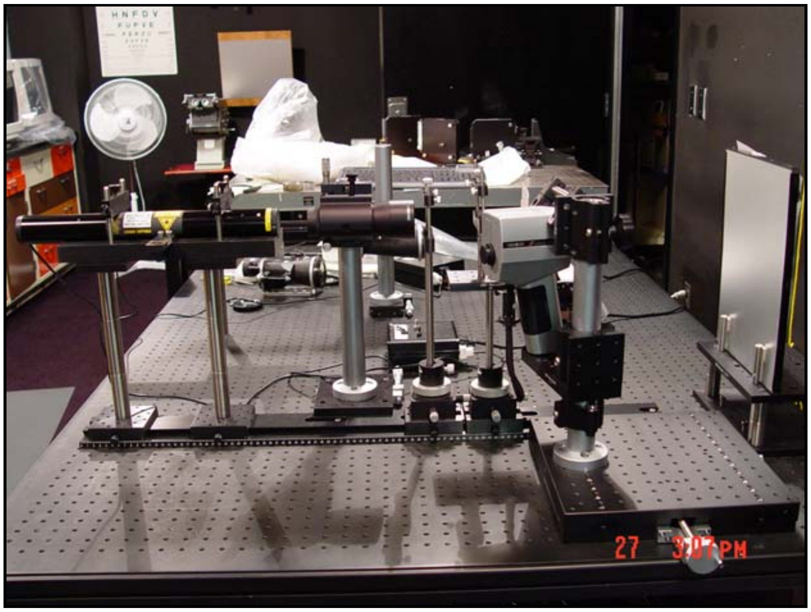


Figure 16. Device for measuring light scatter. One of several lasers with a beam expander was placed along the axis of the lens or orthogonal to the filter. The light passing the filter was reflected from a screen. A photometer with near correcting lens was placed on a linear translation stage allowing reflected luminance measurements to be taken along defined positions on the screen. Luminance measurements were taken both with and without the filter in place. The normalized difference was then used as a measure of scatter.

Light Scatter for Filter 105 C Fog 5

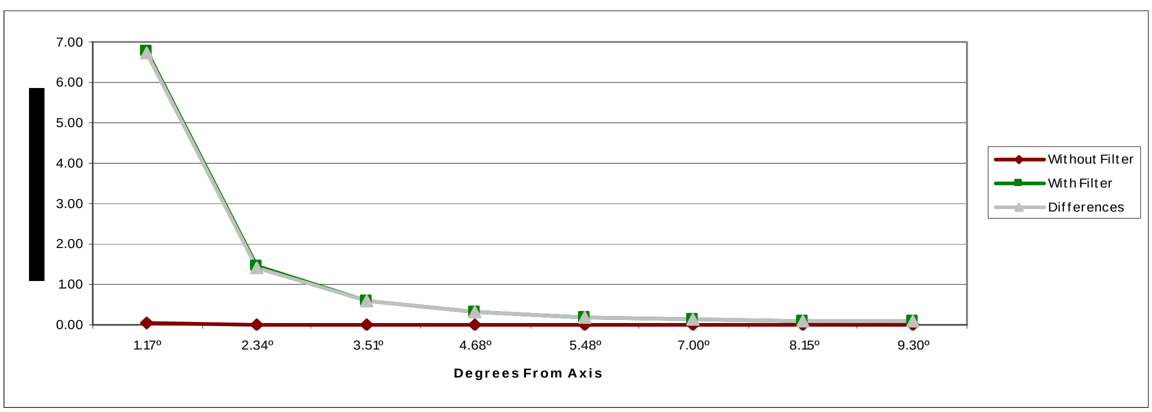


Figure 17. Filter 105 C Fog 5 filter scatter characteristics at wavelength 594.1 nm. Percent of peak luminance was measured against angle for the 105 C Fog 5 filter. The luminance both with and without the filter in place is shown.

The inability to reliably evaluate both blur and scatter using the 2-spot technique led to another approach using the same apparatus; to combine a simple threshold measure of a single spot with the Westheimer and Liang (1995b) procedure for determining scatter (figure 6). The rationale for this hybrid approach was that the peak of the PSF is reduced with even low levels of blur (including blur from aberrations). Once again, a three-phase procedure was used (figure 18). The final psychophysical method developed that was used in the single-spot pilot testing was a 3-reversal staircase (table 3).

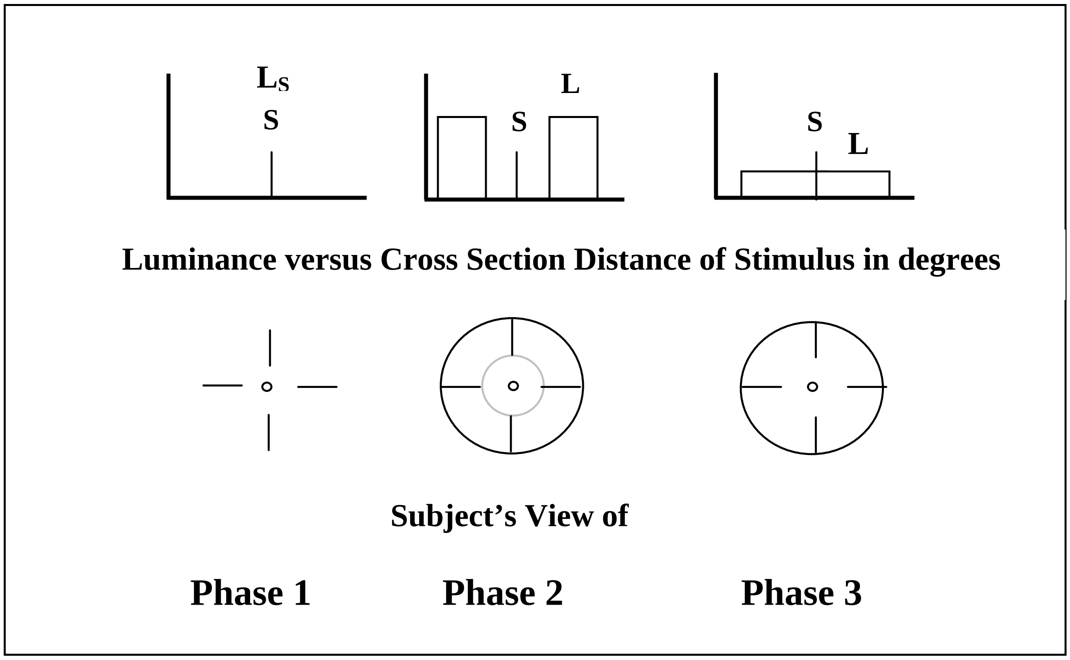


Figure 18. Three-phase trial for glare stimulus presentation. The three stimuli presented during each trial are represented here. During phase 1 the threshold for a spot (S) is determined (Ls) by observer responses; during phase 2 the threshold Luminance of an annulus (L_A) for which S can just be seen is determined by observer responses; and during phase 3 the threshold luminance of a pedestal (L_P) for which S can just be seen is determined by observer responses. The inner diameter of the 20^o annulus in phase 2 is 1^o and the diameter of the pedestal in phase 3 is 20^o. Two black horizontal and two black vertical lines, length equal to the width of the annulus, are always presented as eye alignment guides. A barely visible annulus (just-sufficient to make the lines easily visible) is used in phase 1. During phase 3 the lines are embedded in the 20^o pedestal. Having the observer first determine the single-spot threshold and using this as a basis for the spot luminance in phases 2 and 3 reduced the increment threshold variance.

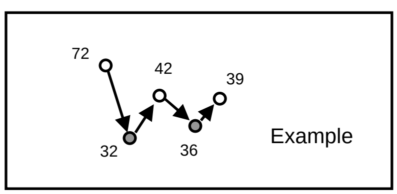
Table 3.

Stair case psychophysical method used with single-spot trials.

I. A three-reversal staircase psychophysical method was employed, requiring a response indicating whether the spot was seen (forced choice). If no determination was made within 2 seconds following a tone cue, the response was interpreted as “not seen.” The spot was automatically initiated, presented, incremented, or terminated following each observer response (timing of spot presentation was semi-randomized). All staircases began and ended with the spot being seen in order to reduce variance by ensuring that the subject saw the spot, knew what it looked like, and knew where it was.

II. In every case, prior to the first reversal, the luminance increments were 10 monitor $\log_{10}$ luminance steps at a time. Following the first reversal, the luminance increments were 5 $\log_{10}$ luminance steps at a time; following the second and third reversals, 3 $\log_{10}$ luminance steps at a time. This increment sequence was designed to minimize variance and decrease testing time. The start luminance for a phase 1 example is diagramed below as the first open circle on the left, 72 monitor $\log_{10}$ luminance steps. The first reversal shown in the example was at 32 monitor $\log_{10}$ luminance steps, the first increment step at which the spot could not be seen. The last increment step at which the spot could just be seen, 39 monitor $\log_{10}$ luminance steps, was recorded as the end point.

Example:
72-40 (4 increments) =32, 32+10 (2 increments) =42, 42-6 (3 increments) =36, 36+3 (1 increment) =39 >>> 39 monitor $\log_{10}$ luminance steps is the end-point



Aviator post refractive surgery pilot data

Toward the end of 2005 an opportunity arose to try out the glare measurement procedure with 8 post refractive surgery pilots. The circumstances precluded getting more than a few minutes with each subject and no reliable history related to night vision could be obtained. However, past records regarding their vision were available, allowing us to relate our measurements to their age, gender, refraction, acuity, haze determination, months post surgery, and wavefront analysis (RMS values). We ran between 3 and 5 successive trials on each subject. This provided at least one practice trial and produced at least one acceptable data collection trial. Criteria for trial rejection are provided in table 4 (appendix C). A 3-phase general glare measurement strategy (figure 18) was used to obtain the data from the 8 post refractive surgery pilots. The psychophysical procedure used is outlined in table 3. The data obtained from the 8 subjects is provided in appendix C.

No sphere correction was employed for these 8 post refractive surgery subjects due to the number of reflecting surfaces required in a trial frame, each of which reduced overall light transmission and produced ghost reflections. No neutral density filters were required during these trials in order to change the luminance range of the device. The spot start luminances refer to the values at the beginning of each stage of the 3-stage procedure employed. For example, the phase 1 single spot start luminance was 72 monitor $\log_{10}$ luminance steps. The purpose of starting at this bright value was to make certain naïve observers new where the spot was and what it looked like, thus reducing data variability and number of rejected trials. The reason for using a barely-visible annulus in phase 1 was to provide two vertical and two horizontal, barely-visible black eye-guides for the subjects. A dark-adapted, sophisticated observer could do the task without eye-guide lines or with an annulus luminance of 4-8 log luminance steps, whereas the naïve subjects found the task very difficult at these low luminances.

The standard deviations of subject responses from phases 1, 2, and 3 data were relatively small, based on our experience with this type of data. As might be expected, the total RMS for these subjects was modestly correlated with their uncorrected log MAR acuity, namely 0.669. The highest correlation found between total RMS and the values obtained using the glare tester was -0.575. It was between total RMS and spot threshold in phase 1. This result may seem counter intuitive, i.e., that luminance required to see a spot with reduced PSF peak would be higher. However, as discussed above, the situation is more complicated than this simple interpretation due to lateral integration effects. Ricco's law holds for very small stimuli in the fovea area (Bartlett, 1966a, Schwartz, 2004), making the spread of light from the peak to the shoulder of the PSF combine to determine the luminance threshold, thereby obscuring the reduction in peak amplitude. Second, this problem is exacerbated by micronystagmus or micro saccades, making foveal fixation problematical in the context of low-luminance spot threshold tasks, particularly for naïve observers. These problems were probably contributors to the poor end-point definition in the two-spot separation task as well. At this stage in developing a glare tester, it is fair to say that the cause of this result is not known and may be chance.

The other correlations were all considerably lower than these. In particular, the correlation between total RMS and the annulus – pedestal luminance was 0.240, suggesting, but by no

means substantiating, the proposition that measurement of forward scatter is poorly related to blur created from higher order aberrations of the eye's optical system.

Discussion

This technical report is a progress report on efforts at USAARL to develop a disability glare tester that can be used to predict night vision problems of post refractive surgery patients, particularly U. S. Army aviators, who generally have good vision. The nature and complexity of the problem was reviewed, along with current methods for assessing disability glare. Current techniques primarily focus on measuring and evaluating the effects of forward scatter in the eye. Computerized versions of the Straylight Meter, probably the best of these approaches, have been very successful and recent efforts to refine procedures for use with naïve subjects have made the technique even better. The central problem is that the forward scatter measurement, while elegant, does not apply well for predicting night vision problems due to aberrations induced by refractive surgery.

There are two approaches that can be taken to deal with this problem. The first is to define disability glare as a forward scatter problem and acknowledge the fact that aberrations induced by refractive surgery are fundamentally different from the mechanisms of forward scatter. The spreading of the PSF due to aberrations in the cornea would justifiably be measured in some different manner. This could be a technique based on some variation of wavefront analysis.

A second approach would be to say that both forward scatter and aberrations produce many of the same visual consequences on a macro scale, particularly reduced contrast. Taking this approach would involve a broader definition of disability glare, one including both intraocular forward light scatter and blur due to aberrations. A measurement procedure would have to cover the consequences of both mechanisms.

There is nothing fundamentally wrong with either approach. Some measures of contrast sensitivity may well turn out to be a very good technique. However, historically, there have been many problems approaching the problem this way, not the least of which is poor reliability. Using the direct compensation or counter-phase flicker method for measuring stray light (Straylight Meter) in combination with a variation of the flat panel diode array method developed by Gutierrez et al. (2003) may prove to be very successful. The research remains to be done.

Here, we tried to develop a hybrid technique, using a variation of the increment threshold method and a simple spot luminance or a 2-spot alternation threshold. There are lateral processing problems, even in central fovea (Ricco's law), along with micronystagmus or micro saccades, that probably cannot be overcome when using a single-spot threshold technique to evaluate the effects of aberrations. However, there was indication that using the individual's single-spot threshold to determine spot luminance when going on to determine increment thresholds reduced data variability. It may well be that an alternating 2-spot threshold procedure in combination with this 3-phase increment threshold procedure may work to provide an

approach for predicting night vision problems in the desired U. S. Army aviator population. Additional work will be needed to determine this.

It may be that using true point sources, would have led to different or less ambiguous results from those that we found. Nonetheless, we opted to capitalize on the flexibility of computer-generated stimuli in this pilot work, important when exploring a variety of options. Computer-generated stimuli can also be highly transportable.

There is much research that can be done in this area. The approaches just discussed and the further development of work started at this lab witness this. Whether one or a combination of the procedures looked at here will prove useful in light of better refractive surgery techniques and more refined and cheaper wavefront analysis devices is a question that time will answer.

Conclusions

The USAARL data presented here is based on ongoing pilot research, not a formal study with a formal experimental design. Different variables and different problems associated with developing a disability glare measurement procedure were evaluated by the researchers in a systematic fashion using brief sets of trials.

So, where does this leave us? Additional work would be necessary to refine the procedures already developed at USAARL. Follow-on studies would be necessary to test whether any of these procedures are actually predictive of night vision problems induced by refractive surgery. Investment in further disability glare tester research is probably justified, but large investments of time and money probably are not. This laboratory does not anticipate any large investiture in time or money for further pursuit of a broadly defined disability glare testing procedure. Refractive surgery techniques are getting better and more refined, having significantly reduced night vision complications. In addition, the refractive surgery studies completed in this laboratory have found that refractive surgery, although having visual consequences, does not significantly reduce a pilot's ability to perform (Bissette, 2006; van de Pol et al., 2007). Similar work in the U. S. Air Force and U. S. Navy has been finding similar results (Baldwin, 2001). The long term consequences of refractive surgery, although not yet known, do not seem to be a problem in the mid-term.

A number of issues and guidelines are apparent from the review above. First, the visual processes in disability glare testing are complicated and dynamic, requiring that a large number of variables be controlled or accounted for. Second, the definition of disability glare probably needs to be expanded to include blur/aberrations, particularly for predicting post refractive-surgery-patient night vision problems. Third, the technique for measuring disability glare should be pupil-sparring over the entire range of measurements if the visual impact of a patient's natural pupil at night is to be assessed. Fourth, a disability glare testing procedure needs to be reliable and sensitive to small differences in intraocular light scatter and higher order aberrations, providing a fairly large measurement range (scale). Fifth, a method must be found for converting raw data to a validated disability glare index scale. Sixth, the disability glare testing

procedure needs to be easy for a patient and relatively quick, particularly if it is going to be used as a clinical or screening tool. If increment threshold measurements are going to be used, notoriously difficult for naïve observers, it needs to be demonstrated that reliable, valid measurements can be made. Seventh, any glare measurement technique that will be used for predicting post refractive surgery night vision problems must be validated using post refractive surgery patients. Objective measures of their night vision performance needs to be assessed as well as their reported night vision problems. Eighth, a disability glare tester should be easy for technicians to use and produce reliably comparable results independent of who is doing the testing. Ninth, the measurement system should be easy to transport if it is to be used as a screening tool. Tenth, disability glare measurement procedures must ensure protections against patient manipulation of result outcomes. This is particularly important if the procedure is going to be used for job-related vision testing or testing of drivers or pilots (De Wit et al., 2005b).

References

Adams, A. J., Brown, B., Haegerstrom-Portnoy, B., Flom, M. C. and Jones, R. T. 1978. Marijuana, alcohol, and combined drug effects on the time course of glare recovery. Psychopharmacology. 56(1): 81-86.

Allen, M. J., Abrams, B. S., Ginsburg, A. P. and Weintraub, L. 2001. Forensic Aspects of Vision and Highway Safety. Lawyers and Judges Publishing Company: Tucson, Arizona.

Anderson, S. J. and Hollidy, I. E. 1995. Night driving: Effects of glare from vehicle headlights on motion perception. Ophthalmology and Physiological Optics. 15: 545-551.

Applegate, R. A. 1989. Disability glare and hydrogel lens wear revisited. Optometry and Vision Science. 66(11): 756-759.

Applegate, R. A., Ballentine, C., Gross, H., Sarver, E. J., and Sarver, C. A. 2003. Visual acuity as a function of Zernike mode and level of root mean square error. Optometry and Vision Science. 80(2): 97-105.

Applegate, R. A., Hilmantel, G., and Thibos, L. N. 2001. Visual Performance Assessment. Chapter seven. In S. M. Macrae, R. R. Krueger and Applegate, R. A. (eds.) Customized Corneal Ablation: The Quest for SuperVision. Slack: Thorofare, New Jersery: 81-92.

Applegate, R. A., Trick, L. R., Meade, D. L., and Hartstein, J. 1987. Radial keratotomy increases the effects of disability glare: Initial results. Annals of Ophthalmology. 19: 293-297.

Applegate, R. A. and Wolf, M. 1987. Disability glare increased by hydrogel lens wear. American Journal of Optometry and Physiological Optics. 64(5): 309-312.

Artal, P., Berrio, E., and Guirao, A. 2002. Contribution of the cornea and internal surfaces to the change of ocular aberrations with age. Optical Society of America. 19(1): 137-143.

Artal, P, Guirao, A, Berrio, E, Piers, P., and Norrby, S. Spring 2003. Optical aberrations and the aging eye. International Ophthalmology Clinics. 43(2): 63-77.

Artal, P., Guirao, A., Berrio, E., and Williams, D. R. 2001. Compensation of corneal aberrations by the internal optics of the eye. Journal of Vision. 1(1): 1-8. http://journalofvision.org

Atchison, D. A. and Smith, G. 2000. Optics of the Human Eye. Butterworth-Heinemann: Oxford.

Bailey, M. D., Mitchell, G. L., Dhaliwal, D. K., Boxer-Wachler, B. S., and Zadnik, K. 2003. Patient satisfaction and visual symptoms after laser in situ keratomileusis. Ophthalmology. 110(7): 1371-1378.

Baker, H. D. 1949. The course of foveal light adaptation measured by the threshold intensity increment. Journal of the Optical Society of America. 39(2): 172-179.

Baker, H. D. 1953. The instantaneous threshold and early dark adaptation. Journal of the Optical Society of America. 43(9): 798-803.

Baldwin, J. B. (personal communication). 2001. Including the Modified Mesotest II and Straylight Meter testing procedures. Brooks Air Force Base, San Antonio, Texas.

Barbur, J.L. 2004 (personal correspondence)

Barlow, H. B. 1972. Dark and light adaptation: Psychophysics (Chapter 1). In D. Jameson and L.M. Hurvich (eds.), Visual Psychophysics. Springer-Verlag: New York. 1-28.

Bartlett, N.R. 1966a. Thresholds as dependent on some energy relations and characteristics of the subject (Chapter 7). In Clarence H. Graham (ed.), Vision and Visual Perception. John Wiley and Sons: New York. 154-184.

Bartlett, N.R. 1966b. Dark adaptation and light adaptation (Chapter 8). In Clarence H. Graham (ed), Vision and Visual Perception. John Wiley and Sons: New York. 185-207.

Bartlett, J. D. and Jaanus, S. D. (eds.). 1995. Clinical Ocular Pharmacology. Third Edition. Butterworth-Heinemann: Boston.

Berman, S. M., Fein, G., Jewett, D. L., and Ashford, F. 1993. Luminance-controlled pupil size affects Landolt C task performance. Journal of the Illuminating Engineering Society. The 1992 IESNA Annual Conference. 150-165.

Berman, S. M., Fein, G., Jewett, D. L., and Ashford, F. Summer 1994. Landolt-C recognition in elderly subjects is affected by scotopic intensity of surround illuminants. Summer 1994. Journal of the Illuminating Engineering Society. The 1993 IESNA Annual Conference. 123-128.

Berman, S. M., Jewett, D. L., Fein, G., Benson, B., Law, T., Myers, A., and Bullimore, M. A. Winter 1996. Lighting spectral effect on Landolt C performance is enhanced by blur and abolished by mydriasis. Journal of the Illuminating Engineering Society. The 1995 IESNA Annual Conference. 42-50.

Bichao, I. C., Yager, D., and Meng, J. 1995. Disability glare: Effects of temporal characteristics of the glare source and of the visual field location of the test stimulus. Journal of the Optical Society of America. A12(10): 2252-2258.

Bissette, G. M. 2006. Personal communication regarding the Evaluation of Refractive Surgery for Army Aviation and the Operational Assessment of Refractive Surgery for Rated Army Aviators: A Prospective Evaluation studies. U. S. Army Aeromedical Research Laboratory, Fort Rucker, Alabama.

Blaho, K. E., Connor, C. G., and Winbery, S. L. 1998. Pharmacology and refraction. In William J. Benjamin (ed.), Borish's Clinical Refraction. W. B. Saunders Company: Philadelphia. 369-407.

Boynton, R. M., Bush, W. R,. and Enoch, J. M. 1954. Rapid changes in foveal sensitivity resulting from direct and indirect adapting stimuli. Journal of the Optical Society of America. 44(1): 56-60.

Bron, A. J., Tripathi, R. C., and Tripathi, B. J. 1997. Wolff's Anatomy of the Eye and Orbit. 8th Edition. Chapman and Hall Medical: London.

Brown, J.L. 1966. Afterimages (Chapter 17). In C. H Graham (ed.), Vision and Visual Perception. John Wiley & Sons: Inc. New York. 479-503.

Brunette, I, Gresset, J., Boivin, J., Pop, M, Thompson, P, Lafond, G. P., and Makni, H. 2000. Functional outcome and satisfaction after photorefractive keratectomy. Ophthalmology. 107(9): 1790-1796.

Campbell, C. E. 2003. A new method for describing the aberrations of the eye using Zernike polynomials. Optometry and Vision Science. 80(1): 79-83.

Chalita, M. R., Chavala, S., Xu, M., and Krueger, R. R. 2004. Wavefront analysis in post-LASIK eyes and its correlation with visual symptoms, refraction, and topography. Ophthalmology. 111(3): 447-453.

Chalita, M. R.and Krueger, R. R. 2004. Correlation of aberrations with visual acuity and symptoms. Ophthalmology Clinics of North America. 17: 135-142.

Chisholm, C. M., Evans, A. D. B., Harlow, J. A., and Barbur, J. L. 2003. New test to assess pilot's vision following refractive surgery. Aviation, Space, and Environmental Medicine. 74(5): 551-559.

Chou, B. and Wachler, B. S. B. 2001. The role of pupil size in refractive surgery. Refractive Source. http://www.refractivesource.com/

Cox, M. J., Norman, J. H., and Norman, P. 1999. The effect of surround luminance on measurements of contrast sensitivity. Ophthalmology and Physiological Optics. 19(5): 401-414.

Ciuffreda, K. J. 1998. Accommodation, the pupil, and presbyopia. Chapter 4. In W. J. Benjamin (ed.), Borish's Clinical Refraction. W. B. Saunders Company: Philadelphia. 77-120.

DeMott, D. W. and Boynton, R. M. 1958. Sources of entoptic stray light. Journal of the Optical Society of America. 48(2): 120-125.

De Waard. P. W. T., IJspeert, J. K., van den Berg, T. J. T. P., and de Jong, T. V. M. 1992. Intraocular light scattering in age-related cataracts. Investigative Ophthalmology and Visual Science. 33(3): 618-625.

De Wit, G. C. 2002. Development of a stray light meter for driver licensing. Chapter 1. In EU project: 1-TREN E3 200/7/S12.282826. In T . J. T. P. van den Berg (Project Leader) Assessment of Visual Function of Driving-License Holders. 9-16. www.glare.be/EUreport 2002%20.pdf

De Wit, G. C. and Coppens, J. E. 2003. Stray light of spectacle lenses compared with stray light in the eye. Optometry and Vision Science. 80(5): 395-400.

De Wit, G. C., Franssen, L., Coppens, J. E., and van den Berg, T. J. T. P. 2005a. Glare sensitivity and visual impairment. Chapter 3. In van den Berg, T. J. T. P. and van Rijn, L. J. (Project Leaders). 2005. Relevance of Glare Sensitivity and Impairment of Visual Function Among European Drivers (European Glare Study). final report EU project SUB-B27020B-E3-GLARE-2002-S07.180. 77-96. http://www.glare.be/publications.htm

De Wit, G. C., Franssen, L., Coppens, J. E., and van den Berg, T. J. T. P. 2005b. Measuring glare sensitivity. Chapter 2. In van den Berg, T. J. T. P. and van Rijn, L. J. (Project Leaders). 2005. Relevance of Glare Sensitivity and Impairment of Visual Function Among European Drivers (European Glare Study). final report EU project SUB-B27020B-E3-GLARE-2002-S07.180. 53-76. http://www.glare.be/publications.htm

Elliott, D.B. 1998. Contrast sensitivity and glare testing (Chapter 8). In W. J. Benjamin (ed.), Borish's Clinical Refraction. W. B Saunders Company: Philadelphia, Pennsylvania. 203-241.

Elliott, D. B. and Bullimore, M. A. 1993. Assessing the reliability, discriminative ability and validity of disability glare tests. Investigative Ophthalmology & Visual Science. 34(1): 108-119.

Elliott, D.B., Mitchell, S., and Whitaker, D. 1991. Factors affecting light scatter in contact lens wearers. Optometry and Vision Science. 68(8): 629-633.

Elliott, D. B., Sanderson, K., and Conkey, A. 1990. The reliability of the Pelli-Robson contrast sensitivity chart. Ophthalmic and Physiological Optics. 10(1): 21-24.

Fan-Paul, N. I., Li, J., Miller, J. S., and Florakis, G. J. 2002. Night vision disturbances after corneal refractive surgery. Survey of Ophthalmology. 47(6): 533-546.

Franssen, L., Coppens, J. E., and van den Berg, T. J. T. P. 2006. Compensation comparison method for assessment of retinal straylight. Investigative Ophthalmology & Visual Science. 47(2): 768-776.

Fry, G. A. and Alpern, M. 1953. The effect of a peripheral glare source upon the apparent brightness of an object. Journal of the Optical Society of America. 43(3): 189-195.

Ghaith, A. A., Daniel, J., Stulting, R. D., Thompson, K. P., and Lynn, M. 1998. Contrast sensitivity and glare disability after radial keratotomy and photorefractive keratectomy. Archives of Ophthalmology. 116(1): 12-18.

Graham, C. H. 1966a. Some basic terms and methods (Chapter 3). In C. H Graham (ed.), Vision and Visual Perception. John Wiley & Sons, Inc.: New York. 60-67.

Graham, C. H. 1966b. Some fundamental data (Chapter 4). In C. H. Graham (ed.), Vision and Visual Perception. John Wiley & Sons, Inc.: New York. 68-80.

Green, M. 2004. Seeing Pedestrians at Night. www.visualexpert.com/Resources/pedestrians.html

Green, M. and Senders, J. 2004. Human Error in Road Accidents. www.visualexpert.com/Resources/roadaccidents.html

Guitierrez, R., Jimenez, J. R., Villa, C., Valverde, J. A., and Anera, R. G. 2003. A simple device for quantifying the effects of halos after ocular surgery. Journal of Biomedical Optics. 4: 663-667.

Haegerstrom-Portnoy, G., Schneck, M. E., and Brabyn, J. A. 1999. Seeing into old age: vision function beyond acuity. Optometry and Vision Science. 76(3): 141-158.

Harrison, J.M., Tennant, T.B., Gwin, M.C., Applegate, R.A., Tennant, J.L., van den Berg, T.J.T.P., and Lohmann, C.P. 1995. Forward light scatter at one month after photorefractive keratectomy. Journal of Refractive Surgery. 11: 83-88.

Hecht, E. and Aajac, A. 1987. Optics. Second Edition. Addison-Wesley: Reading, Massachusetts.

Hemenger, R. P. 1990. Light scatter in cataractous lenses. Ophthalmological Physiological Optics. 10: 394-397.

Holladay, L. L. 1926. The fundamentals of glare and visibility. Journal of the Optical Society of America. 12(2): 271-319.

Holladay, L. L. 1927. Action of a light-source in the field of view in lowering visibility. Journal of the Optical Society of America. 14(1): 1-15.

IJspeert, J. K., De Waard, P. W. T., van den Berg, T. J. T. P., and De Jong, P. T. V. M. 1990. The intraocular straylight function in 129 healthy volunteers; dependence on angle, age and pigmentation. Vision Research. 30(5): 699-707.

Jaanus, S. D. 1995. Anti-edema drugs. Chapter 15. In J. D. Bartlett and S. D. Jaanus (eds.), Clinical Ocular Pharmacology. Third Edition. Butterworth-Heinemann: Boston. 369-388.

Jones, L. W. and Jones, D. A. 2000. Common Contact Lens Complications: Their Recognition and Management. Butterworth-Heinemann: Oxford.

Kalich, M. E. 2004. More Reliable Glare Testing. USAARL ILIR study approved 01/15/04. Fort Rucker, Alabama.

Kanski, J. L. 2003. Clinical Ophthalmology: A Systematic Approach. 5th Edition. Butterworth Heinemann: Philadelphia.

Kaufman, P. L. and Alm, A. (ed.s). 2002. Adler's Physiology of the Eye: Clinical Application. 10th Edition. C. V. Mosby: Philadelphia.

Keating, M. P. 1988. Geometrical, Physical, and Visual Optics. Butterworth-Heinemann: Boston.

Kelly, J. E., Mihashi, T., and Howland, H. C. 2004. Compensation of corneal horizontal/vertical astigmatism, lateral coma, and spherical aberration by internal optics of the eye. Journal of Vision, 4(4): 262-271. http://www.journalofvision.org/4/4/2/,doi:10.1167/4.4.2

Klein, B. E., K., Klein, R., and Linton, K. L. P. 1992. Prevelance of age-related lens opacities in a population. Ophthalmology. 99(4): 546-552.

Klein, S. A. 2001. Problems with wavefront aberrations applied to refractive surgery: Developing standards. In Ophthalmic Technologies XI, SPIE Proceedings Volume 4245. SPIE: Bellingham, Washington. 47-56.

Klein, S. A., Hoffmann, S., and Hickenbotham, S. C. 2003. Improved glare (halos and scattered light) measurement for post-LASIK surgery. Mopane 2003. Vision Processing Laboratory, Vision Science Program UC Berkeley. http://cornea.berkeley.edu/presentations/

Klymenko, V. 2001. Personal communication. USAARL, Fort Rucker, Alabama.

Korb, D. R., Craig, J., Doughty, M., Guillon, J., Smith, G., and Tomlinson, A. 2002. The Tear Film: Structure, Function and Clinical Examination. Butterworth-Heinemann: London.

Krizaj, D. 2000. Mesopic state: Cellular mechanisms involved in pre- and post-synaptic mixing of rod and cone signals. Microscopy Research and Technique. 50: 347-359.

Krizaj, D. and Hawlina, M. 2002. Cellular basis for rod-cone interactions in the outer retina. ZDRAV VESTN. 71(Supplement II), 137-145.

Lewis, A. L. 1993. Basic concepts in environmental lighting. Chapter Five. In D. G. Pitts and R. N. Kleinstein (eds.), Environmental Vision: Interactions of the Eye, Vision, and the Environment. Butterworth-Heinemann: Boston. 137-150.

Liang, J. and Westheimer, G. 1995. Optical performances of human eyes derived from double-pass measurements. Journal of the Optical Society of America. 12(7): 1411-1416.

Liang, J., Williams, D. R., and Miller, D. T. 1997. Supernormal vision and high-resolution retinal imaging through adaptive optics. Journal of the Optical Society of America. 14(11): 2884-2892.

Lowenstein, O. and Loewenfeld, I. E. 1969. The pupil. (Chapter 9). In H. Davson (ed.), The Eye. Academic Press: New York. 255-337.

Marsack, J. D., Thibos, L. N., and Applegate, R. A. 2004. Metrics of optical quality derived from wave aberrations predict visual performance. Journal of Vision. 4(4): 322-328. http://journalofvision.org/4/4/8,doi:10.1167/4.4.8

Martinez, C.E., Applegate, R.A., Klyce, S.D., McDonald, M.B., Medina, J.P., and Howland, H.C. 1998. Effect of pupillary dilation on corneal optical aberrations after photorefractive keratectomy. Archives of Ophthalmology. 116: 1053-1062.

Maurice, D. M. 1957. The structure and transparency of the cornea. Journal of Physiology. 136: 263-287.

Melki, S. A., Proano, C. E., and Azar, D. T. 2003. Optical aberrations after LASIK. Chapter 28. D. T. Azar and D. D. Koch (ed.s), LASIK Fundamentals, Surgical Techniques, and Complications. Marcel Dekker: New York. 387-402.

McLeod, S. D. 2001. Beyond Snellen acuity. Archives of Ophthalmology. 119: 1371-1373.

Nakagawara, V. B., Montgomery, R. W., and Wood, K. J. 1994. The applicability of commercial glare test devices in the aeromedical certification of pilot applicants. Final Report DOT/FAA/AM-94/15. U. S. Department of Transportation. Federal Aviation Administration. Civil Aeromedical Institute. Oklahoma City, Oklahoma 73125.

Navarro, R., Moreno, E., and Corronsoro, C. 1998. Monochromatic aberrations and point-spread functions of the human eye across the visual field. Journal of the Optical Society of America. 15(9): 2522-2529.

OCULUS. <u>Mesotest II Instruction Manual (English language version)</u>. OCULUS Optikgerate GmbH, Munchholzhauser Str. 29, D – 35582 Wetzlar.

Oshika, T., Klyce, S. D., Applegate, R. A., Howland, H. C., and El Danasoury, M. A. 1999. Comparison of corneal wavefront aberrations after photorefractive keratectomy and laser in situ keratomileusis. American Journal of Ophthalmology.127(1): 1-7.

Pitts, D. G. 1993. Electromagnetic spectrum. (Chapter 4). In D. G. Pitts and R. N. Kleinstein (eds.), <u>Environmental Vision: Interactions of the Eye, Vision, and the Environment</u>. Butterworth-Heinemann: Boston. 87-135.

Quantock, A. J. December 15, 2000. The cornea is clear – but why? OT, 1-4. http://www.optometry.co.uk/articles/docs/377a0e69e398a201252a0bb3680f7f88_quantock20001215.pdf

Pulling, N. H., Wolf, E., Sturgis, S. P., Vaillancourt, D. R., and Dolliver, J. J. 1980. Headlight glare resistance and driver age. Human Factors. 22(1): 103-112.

Rabin, J. 1995 (January). Small letter contrast sensitivity: An alternative measure of visual resolution for aviation candidates. Aviation, Space, and Environmental Medicine.

Rabin, J. 1994. Luminance effects on visual acuity and small letter contrast sensitivity. Optometry and Vision Science. 71(11): 685-688.

Rosenfield M. 1998. Refractive status of the eye (Chapter 1). In W. J. Benjamin (ed.), <u>Borish's Clinical Refraction</u>. W. B Saunders Company: Philadelphia, Pennsylvania. 2-29.

Rushton, W. A. H. and Gubisch, R. W. 1966. Glare: Its measurement by cone threshold and by the bleaching of cone pigments. Journal of the Optical Society of America. 56(1): 104-110.

Sagawa, K. and Takeichi, K. 1992. System of mesopic photometry for evaluating lights in terms of comparative brightness relationships. Journal of the Optical Society of America. 9(8): 1240-1246.

Schieber, F. 1994a. Age and glare recovery time for low-contrast stimuli. <u>Proceedings of the Human Factors and Ergonomics Society</u>. Santa Monica, California: Human Factors and Ergonomics Society. 496-499.

Schieber, F. 1994b. <u>Recent Developments in Vision, Aging and Driving: 1988-1994</u>. Report No. UMTRI-94-26. Ann Arbor, Michigan: University of Michigan, Transportation Research Institute. <u>www.usd.edu/~schieber/pdf/umtri.pdf</u>

Schieber, F. 1995. *Effects of visual aging upon driving performance. In W. Adrian (ed.), <u>Lighting for aging vision and health</u>.* New York: Lighting Research Institute. 113-128.

Schneck, M. E., Adams, A. J., Huie, K., and Lee, E. 1993. A filter for simulating color and spatial vision of the elderly. In B. Drum (ed.), Colour Vision Deficiencies XI. Kluwer Academic Publishers: Netherlands. 357-363.

Schumer, D. J., Bains, H. S., and Brown, K. L. 2000. Dark-adapted pupil sizes in prospective evaluation of laser in situ keratomileusis patients. Journal of Refractive Surgery. 16: S239-S41.

Schwartz, S. H. 1994. Visual Perception: A Clinical Orientation. Appleton & Lange: Norwalk, Connecticut.

Seiler, T., Kaemmerer, M., Mierdel, P., and Krinke, H. E. 2000. Ocular optical aberrations after photorefractive keratectomy for myopia and myopic astigmatism. Archives of Ophthalmology. 118: 17-21.

Shinsuke, S., Kamitani, Y., and Nishida, S. 2001. Afterimage of perceptually filled-in surface. Science. 293: 1677-1680.

Smith, G. 2002. Disability glare and its clinical significance. Optometry Today-Optics Today. 34-37. http://www.optometry.co.uk/articles/20020419/smith20020418.pdf

Stabell, B. and Stabell, U. 1979. Rod and cone contributions to change in hue with eccentricity. Vision Research. 19: 1121-1125.

Stabell, B. and Stabell, U. 1998. Chromatic rod-cone interaction during dark adaptation. Journal of the Optical Society of America. 15(11): 2809-2815.

Stiles, W. S. 1929. The effect of glare on the brightness difference threshold. Proceedings of the Royal Society (London). B 104: 322-350.

Stiles, W. S. and Crawford, B. H. 1937. The effect of a glaring light source on extrafoveal vision. Proceedings of the Royal Society (London). B 122: 255-280.

Swanson, M. W. 1998. Changes in crystalline lens density with age, and fundus reflections (Chapter 32). In W. J. Benjamin (ed.), Borish's Clinical Refraction. W. B Saunders Company: Philadelphia. 1194-1210.

Thibos, L. N. 2001. Assessment of Optical quality (Chapter Six). In S. M. Macrae, R. R. Krueger and Applegate, R. A. (Eds.) Customized Corneal Ablation: The Quest for SuperVision. Slack: Thorofare, New Jersery. 67-78.

Thibos, L. N., Applegate, R. A., Schwiegerling, J. T., Webb, R., and VSIA Task Force Members. 2002. Standards for reporting the optical aberrations of the eye. Journal of Refractive Surgery. 18: S652-S660.

Thibos, L. N., Hong, X., Bradley, A., and Cheng, X. 2002. Statistical variation of aberration structure and image quality in a normal population of healthy eyes. Journal of the Optical Society of America. 19(12): 2329-2348.

Thomson, D. 2001. Methods of assessing cataract and the effect of opacities on vision. Optometry Today-Optics Today. 26-30. http://www.optometry.co.uk/articles/20010601/thomson.pdf

van den Berg, T. J. T. P. 1986. Importance of pathological intraocular light scatter for visual disability. Documenta Ophthalmologica. 61: 327-333.

van den Berg, T. J. T. P. 1991. On the relation between glare and straylight. Documenta Ophthalmologica. 78: 177-181.

van den Berg, T. J. T. P. 1994. On the relation between intraocular straylight and visual function parameters. Investigative Ophthalmology & Visual Science. 35(6): 2659-2660.

van den Berg, T. J. T. P. 1995. Analysis of intraocular straylight, especially in relation to age. Optometry and Vision Science. 72(2): 52-59.

van den Berg, T.J.T.P. (Project Leader). 2002. <u>Assessment of Visual Function of Driving-Licence Holders</u>, final report EU project I-TREN E3 200/7/SI2.282826 (2003). http://www.glare.be/publications.htm

van den Berg, T. J. T. P. and Boltjes, B. 1988. The point-spread function of the eye from 0^o to 100^o and the pattern electroretinogram. Documenta Ophthalmologica. 67: 347-354.

van den Berg, T. J. T. P., de Waard, P. W. T., IJspeert, J. K., and de Jong, P. T. V. M. 1989. Intraocular light scattering assessed quantitatively in age-related cataract. Investigative Ophthalmology & Visual Science. 30 (Suppl.): 499.

van den Berg, T. J. T. P., IJspeert, J. K., and de Waard, P. W. T. 1991. Dependence of intraocular straylight on pigmentation and light transmission through the ocular wall. Vision Research. 31(7/8): 1361-1367.

van den Berg, T. J. T. P. and van Rijn, L. J. (Project Leaders). 2005. <u>Relevance of Glare Sensitivity and Impairment of Visual Function Among European Drivers (European Glare Study)</u>. final report EU project SUB-B27020B-E3-GLARE-2002-S07.180. http://www.glare.be/publications.htm

van de Pol, C. June 7, 2000. Evaluation of refractive surgery for army aviation. USAARL Protocol, HSRRB log # A-9923, HUC log # 157.

van de Pol, C. 2001. Personal communication. USAARL, Fort Rucker, Alabama.

van de Pol, C. 2004. Personal correspondence. Navy Refractive Surgery Center, San Diego, California.

van de Pol, C., Bower, K. S., and Rabin, J. C. December 5, 2001. Operational assessment of refractive surgery for rated aviators. USAARL Protocol, HSRRB log # A10105.2, HUC log # 189.

van de Pol, C., Greig, J. L., Estrada, A., Bissette, G. M., and Bower, K. S. 2007. Visual and flight performance recovery after PRK or LASIK in helicopter pilots. Aviation, Space, and Environmental Medicine 78(6): 547-553.

van de Pol, C., Soya, K., and Hwang, D. G. 2001. Objective assessment of transient corneal haze and its relation to visual performance after photorefractive keratectomy. American Journal of Ophthalmology 132 (2): 204-210.

van Rijn, L. J. 2002. Clinical studies on straylight and glare. In EU project: 1-TREN E3 200/7/S12.282826. Project Leaders T. J. T. P. van den Berg. Assessment of Visual Function of Driving-License Holders. 17-52. www.glare.be/EU_report 2002%20.pdf

van Rijn, L. J., Nischler, C., Gamer, D., Franssen, L., de Wit, G., Koper, R., Vanhoff, D., Grabner, G., Wilhelm, H., Volker-Dieben, H. J., and van den Berg, T. J. T. P. 2005. Measurement of stray light and glare: comparison of Nyktotest, Mesotest, straylight meter, and computer implemented stray light meter. British Journal of Ophthalmology. 89: 345-351. http://bjo.bmjjournals.com

van Rijn, L. J., Wilhelm, H., Emesz, M., Kaper, R., Heine, S., Nitsch, S., Grabner, G., and Volker-Dieben, H. J. 2005. Relation between perceived driving disability and score of vision screening tests. British Journal of Ophthalmology. 86, 1262-1264. http://bjo.bmjjournals.com

Vos, J. J. 1963. Contribution of the fundus oculi to entoptic scatter. Journal of the Optical Society of America. 53: 1449-1451.

Vos, J. J. 2003. On the cause of disability glare and its dependence on glare angle, age and ocular pigmentation. Clinical and Experimental Optometry (Journal of the Australian Optometrical Association). 86(6): 363-370.

Vos, J. J. and Boogaard, J. 1963 Contribution of the cornea to entoptic scatter. Journal of the Optical Society of America. 53: 869-873.

Vos, J. J. and Bouman, M. A. 1964. Contribution of the retina to entoptic scatter. Journal of the Optical Society of America. 54: 95-101.

Vos, J.J. and van den Berg, T.J.T.P. 1999. Report on disability glare. In CIE Collection 1999: Vision and Colour. Physical Measurement of Light and Radiation. CIE 135-1999. 1-9.

Wachler, B. S. B., Durrie, D. S., Assil, K. K., and Krueger, R. R. 1999. Improvement of visual function with glare testing after photorefractive keratectomy and radial keratotomy. American Journal of Ophthalmology. 128(5): 582-587.

Westheimer, G. 1967. Spatial interaction in human cone vision. Journal of Physiology (London). 190: 139-154.

Westheimer, G. and Liang, J. 1995a. Evaluating diffusion of light in the eye by objective means. Investigative Ophthalmology and Visual Science. 35(6): 2652-2657.

Westheimer, G. and Liang, J. 1995b. Influence of ocular light scatter on the eye's optical performance. Journal of the Optical Society of America. 12(7): 1417-1424.

Whitaker, D., Steen, R., and Elliott, D. B. 1994. The authors reply to T. J. T. P. van den Berg's note to the editor, "on the relation between intraocular straylight and visual function parameters." Investigative Ophthalmology and Visual Science. 35(6): 2660-2661.

Wooten, B. R. and Geri, G. A. 1987. Psychophysical determination of intraocular light scatter as a function of wavelength. Vision Research. 27(8): 1291-1298.

Appendix A.

Monitor log luminance steps.

A maximum of seventy-two 0.05 $\log_{10}$ cd/m^2 steps from -1.309 to +1.291

Monitor $\log_{10}$ Luminance Step Increment	Incremented log luminance (Log_{10} cd/m^2)	Luminance (cd/m$^{2)}$	Closest Monitor Brightness Step
0	-1.691	0.02	0
1	-1.641	0.02	0
2	-1.591	0.03	0
3	-1.541	0.03	0
4	-1.491	0.03	1
5	-1.441	0.04	2
6	-1.391	0.04	3
7	-1.341	0.05	3
8	-1.291	0.05	4
9	-1.241	0.06	5
10	-1.191	0.06	6
11	-1.141	0.07	7
12	-1.091	0.08	7
13	-1.041	0.09	8
14	-0.991	0.10	9
15	-0.941	0.11	10
16	-0.891	0.13	11
17	-0.841	0.14	11
18	-0.791	0.16	12
19	-0.741	0.18	13
20	-0.691	0.20	14
21	-0.641	0.23	15
22	-0.591	0.26	16
23	-0.541	0.29	18
24	-0.491	0.32	19
25	-0.441	0.36	20
26	-0.391	0.41	22
27	-0.341	0.46	23
28	-0.291	0.51	25
29	-0.241	0.57	26
30	-0.191	0.64	28
31	-0.141	0.72	30
32	-0.091	0.81	31
33	-0.041	0.91	34
34	0.009	1.02	35
35	0.059	1.15	37
36	0.109	1.29	39

Monitor log_{10} Luminance Step Increment	Incremented log luminance (Log_{10} cd/m^2)	Luminance ($cd/m^{2)}$	Closest Monitor Brightness Step
37	0.159	1.44	41
38	0.209	1.62	43
39	0.259	1.82	46
40	0.309	2.04	48
41	0.359	2.29	51
42	0.409	2.56	53
43	0.459	2.88	56
44	0.509	3.23	59
45	0.559	3.62	62
46	0.609	4.06	66
47	0.659	4.56	69
48	0.709	5.12	73
49	0.759	5.74	76
50	0.809	6.44	80
51	0.859	7.23	84
52	0.909	8.11	88
53	0.959	9.10	93
54	1.009	10.21	98
55	1.059	11.45	103
56	1.109	12.85	109
57	1.159	14.42	114
58	1.209	16.18	120
59	1.259	18.15	126
60	1.309	20.37	132
61	1.359	22.85	139
62	1.409	25.64	147
63	1.459	28.77	155
64	1.509	32.28	162
65	1.559	36.22	171
66	1.609	40.64	180
67	1.659	45.60	189
68	1.709	51.16	199
69	1.759	57.41	210
70	1.809	64.41	221
71	1.859	72.27	233
72	1.909	81.09	245

Appendix B.

Example of 2-spot pilot data.

Example alternating 2-spot trials, from a single researcher observer is presented in table B-1. A constant spot luminance for phases 2 and 3 (figure 11.) was used. Two spot alternation frequencies and 10% and 50% duty cycles were used. Mean annulus and pedestal log luminances at which the alternation was just detected are shown along with standard deviations. The annulus luminance standard deviations were the lowest for the 2-spot trials tried. However, the 2-spot technique did not demonstrate particular sensitivity to different magnitudes of blur, scatter or higher order aberrations (sphere, coma, trefoil). Using a reduced spot size may make this technique more sensitive, but this was not tried.

Table B-1.
Alternating two-spot trials having a constant spot luminance for all three phases.

Subject	Date	# Trials	Annulus Outer Diameter	Annulus Inner Diameter	Double Spot Size (pixels)	Double Spot Separation (pixels)	Double Spot Alternation Frequency	Double Spot Duty Cycle (% time-on)	Constant Double Spot Luminance (monitor log luminance steps)
2	07/29/04 to 08/02/04	21	20^{o}	1^{o}	1	1	1Hz	10%	32
2	07/30/04 to 08/02/04	9	20^{o}	1^{o}	1	1	1Hz	50%	32
2	07/29/04 to 08/02/04	12	20^{o}	1^{o}	1	1	2Hz	50%	32

Mean Annulus Luminance (monitor log luminance steps)	SD Annulus Luminance (monitor log luminance steps)	Mean Pedestal Luminance (monitor log luminance steps)	SD Pedestal Luminance (monitor log luminance steps)	Mean Annulus Luminance - Pedestal Luminance (monitor log luminance steps)	SD Mean Annulus Luminance - Mean Pedestal Luminance (monitor log luminance steps)
37.1	4.4	29.6	6.2	7.5	3.8
51.0	4.6	28.1	5.1	22.9	5.9
50.3	7.2	29.1	5.3	21.2	4.6

Appendix C.

Aviator post refractive surgery pilot data.

Table C-1.
Post refractive surgery pilot trial rejection criteria.

Subject	Run Date	Phase 1 Single Spot Luminance (monitor log luminance steps)	Phase 2 Annulus Luminance (monitor log luminance steps)	Phase 3 Pedestal Luminance (monitor log luminance steps)	COMMENT
RSL 0452	08/30/05	35	72	35	luminance ceiling
RSL 0435	08/31/05	38	57	58	high pedestal luminance
RSL 0080	08/31/05	42	53	11	low pedestal luminance
RSL 0547	08/31/05	44	59	0	No phase 3 response
RSL 0673	08/31/05	46	48	48	annulus and pedestal equal
RSL 0603	08/31/05	14	60	34	low single spot luminance

Description and comments on the following table and graph headings that may not be self-evident:

a. The *Phase 1 Single Spot Luminance* was the phase 1 threshold luminance of the white, 1-pixel spot seen at an annulus center of 0 monitor log luminance steps and an annulus of 11 monitor log luminance steps.
b. The *Phase 2 Annulus Luminance* was the monitor log luminance step at which the center, single spot was just seen by the subject. The central spot during phase 2 was at the phase 1 log luminance threshold plus a predetermined log luminance increase.
c. The *Phase 3 Pedestal Luminance* was the pedestal log luminance step at which the same spot as seen in phase 2 is just seen.
d. *OD Total Post Surgery RMS* was the overall total aberrations of the subject's right eye optics (providing the magnitude of optical surface variations of known shape) measured by wavefront analysis.
e. The only *OD In-device Correction* used with these subjects was an astigmatism correction.
f. *The Dark Pupil Diameter* was the diameter of the right (OD) eye pupil in the dark.
g. *The Maximum Pupil Luminance* gives the maximum monitor log luminance step, of a homogeneous white monitor screen, that does not constrict the OS (left) pupil diameter below 5 mm.
h. The *Last Stair case Increment Size* refers to the size of each stair-case-increment used in phases 2 and 3 (the number of monitor log luminance steps per psychophysical increment).

i. The single spot luminance increase, in log luminance steps, for phases 2 and 3 was shown under the heading, *Phase 2 Single Spot Luminance Increase*. This predetermined value was set to insure that the subject clearly saw the spot.
j. *The Phase 1 Unseen Spot Flag* referred to a trial in which the spot was never seen. This information was used in trial rejections.
k. *The Phase 1 (2 and 3) Non-Response Flag(s)* referred to the number of response cues the subject did not respond to, whether seen or not seen.
l. *The Phase 1 (2 and 3) 5 mm-Pupil Flag* showed whether the pupil dropped below 5 mm at any time during the trial. If it did, the trial was rejected.
m. *Untimely Responses* were responses made outside the response acceptance period of 2 seconds following a response cue.
n. The *Debug Window* indicated whether the program input or output characteristics required calibration.

<u>Table C-2.</u>

Accepted data from post refractive surgery pilots.

Subject	Run Date	Phase 1 Single Spot Luminance (monitor log luminance steps)	Phase 2 Annulus Luminance (monitor log luminance steps)	Phase 3 Pedestal Luminance (monitor log luminance steps)	Annulus Luminance - Pedestal Luminance (monitor log luminance steps
RSL 0248	09/15/05	34.0	54.0	35.0	19.0
RSL 0399	09/15/05	39.0	49.5	31.5	18.0
RSL 0316	09/16/05	44.0	58.5	36.5	22.0
RSL 0305	09/16/05	46.7	59.0	43.3	15.7
RSL 0396	09/19/05	32.5	40.0	23.5	16.5
RSL 0342	09/20/05	41.5	53.0	32.0	21.0
RSL 0407	09/21/05	48.5	49.5	35.0	14.5
RSL 0304	09/26/05	44.5	55.5	29.5	26.0

Pedestal Luminance / Annulus Luminance	Subject Gender	Subject Age	Subject Status	Subject Occupation	Eye Used	Eye Color	Months Post Surgery	OD Total Post Surgery RMS
0.65	Male	24	naïve	Pilot	OD	Hazel	27	0.25
0.64	Male	29	naïve	Pilot	OD	Blue	17	0.13
0.62	Male	31	naïve	Pilot	OD	Light Brown	29	0.27
0.73	Male	30	naïve	Pilot	OD	Blue	25	0.13
0.59	Male	25	naïve	Pilot	OD	Blue	30	0.23
0.60	Male	28	naïve	Pilot	OD	Dark Brown	40	0.20
0.71	Male	23	naïve	Pilot	OD	Dark Brown	50	0.12
0.53	Male	32	naïve	Pilot	OD	Hazel	30	0.14

Viewing Distance	Type Run	ND Filter	Target Guides On Y/N	Last Staircase Increment Size (monitor log luminance steps)	Background Luminance	Annulus Outer Diameter	Annulus Inner Diameter
350	Experimental	None	Yes	3	0	20^o	1^o
350	Experimental	None	Yes	3	0	20^o	1^o
350	Experimental	None	Yes	3	0	20^o	1^o
350	Experimental	None	Yes	3	0	20^o	1^o
350	Experimental	None	Yes	3	0	20^o	1^o
350	Experimental	None	Yes	3	0	20^o	1^o
350	Experimental	None	Yes	3	0	20^o	1^o
350	Experimental	None	Yes	3	0	20^o	1^o

OD Post Surgery Manifest Refraction	OD Snellen Acuity	OD log MAR Acuity	Eyeglasses Y/N	Contact Lenses Y/N	OD In-device Correction	Dark Pupil Diameter	Maximum Pupil Luminance (monitor log luminance steps)
-0.50 -0.00 X 000	20/63	0.5	No	No	none	6	71
0.00 -0.25 X 105	20/40	0.3	No	No	astigmatism	6	71
0.75 -0.75 X 031	20/63	0.5	No	No	astigmatism	6	71
-0.25 -1.00 X 120	20/40	0.3	No	No	astigmatism	6	71
0.00 -0.75 X 175	20/40	0.3	No	No	astigmatism	5	71
-0.50 -0.50 X 095	30/32	0.2	No	No	astigmatism	6	71
0.00 -0.75 X 75	20/40	0.3	No	No	astigmatism	6	71
-0.25 -0.75 X 100	20/32	0.2	No	No	astigmatism	7	71

Single Spot Size (pixels)	Single Spot Color	Single Spot Duration (msec on / msec off)	Phase 1 Adaptation Screen Duration (seconds)	Phase 1 Adaptation Screen Color	Phase 1 Adaptation Screen Luminance (monitor log luminance steps)	Phase 1 Annulus Color	Phase 1 Annulus Luminance (monitor log luminance Steps)
1	W	100 / 900	30	W	15	W	11
1	W	100 / 900	30	W	15	W	11
1	W	100 / 900	30	W	15	W	11
1	W	100 / 900	30	W	15	W	11
1	W	100 / 900	30	W	15	W	11
1	W	100 / 900	30	W	15	W	11
1	W	100 / 900	30	W	15	W	11
1	W	100 / 900	30	W	15	W	11

Phase 1 Center Pedestal Luminance (monitor log luminance steps)	Phase 1 Single Spot Start Luminance (monitor log luminance steps)	Phase 2 Adaptation Screen Duration (seconds)	Phase 2 Adaptation Screen Color	Phase 2 Adaptation Screen Luminance (monitor log luminance steps)	Phase 2 Annulus Color
0	72	30	W	15	W
0	72	30	W	15	W
0	72	30	W	15	W
0	72	30	W	15	W
0	72	30	W	15	W
0	72	30	W	15	W
0	72	30	W	15	W
0	72	30	W	15	W

Phase 2 Annulus Start Luminance (monitor log luminance steps)	Phase 2 Center Pedestal Luminance (monitor log luminance steps)	Phase 2 Single Spot Luminance Increase over Phase 1 Threshold (monitor log luminance steps)	Phase 3 Adaptation Screen Duration (seconds)	Phase 3 Adaptation Screen Color	Phase 3 Adaptation Screen Luminance (monitor log luminance steps)
11	0	2	30	W	15
11	0	2	30	W	15
11	0	2	30	W	15
11	0	2	30	W	15
11	0	2	30	W	15
11	0	2	30	W	15
11	0	2	30	W	15
11	0	2	30	W	15

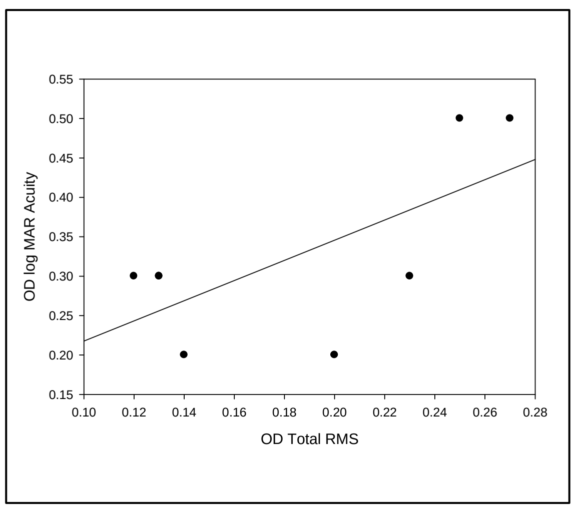
Phase 3 Pedestal Diameter	Phase 3 Pedestal Start Luminance (monitor log luminance steps)	Phase 1 Unseen-Spot Flag	Phase 1 Non-responses	Phase 2 Non-responses	Phase 3 Non-responses	Phase 1 5mm-Pupil Flag (Y/N)	Phase 2 5mm-Pupil Flag (Y/N)	Phase 3 5mm-Pupil Flag (Y/N)
20^{o}	8							
20^{o}	8							
20^{o}	8							
20^{o}	8							
20^{o}	8							
20^{o}	8							
20^{o}	8							
20^{o}	8							

Table C-3.
Standard Deviations from Phases 1, 2, 3.

	Phase 1 Mean & Standard Deviation Single Spot Luminance (monitor log luminance steps)	Phase 2 Mean & Standard Deviation & SD Annulus Luminance (monitor log luminance steps)	Phase 3 Mean & Standard Deviation Pedestal Luminance (monitor log luminance steps)	Mean & Standard Deviation Annulus Luminance - Pedestal Luminance (monitor log luminance steps
Mean	41.34	52.38	33.29	19.09
Standard Deviation	5.79	6.14	5.75	3.79

Table C-4.
Total RMS/Acuity Correlation Data.

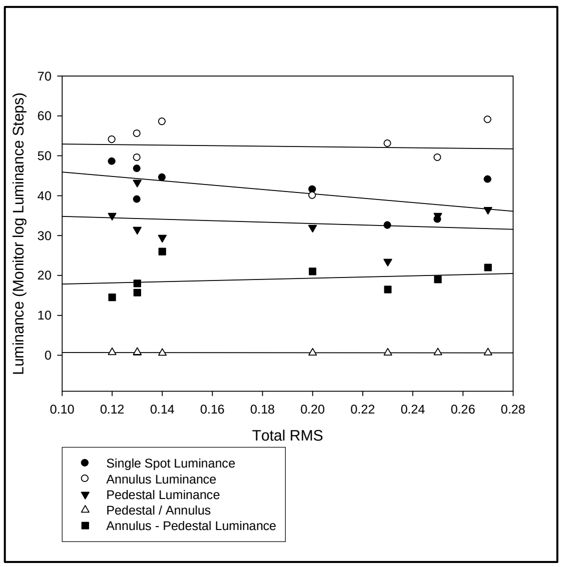
OD Total RMS	OD log MAR Acuity
0.25	0.5
0.13	0.3
0.27	0.5
0.13	0.3
0.23	0.3
0.20	0.2
0.12	0.3

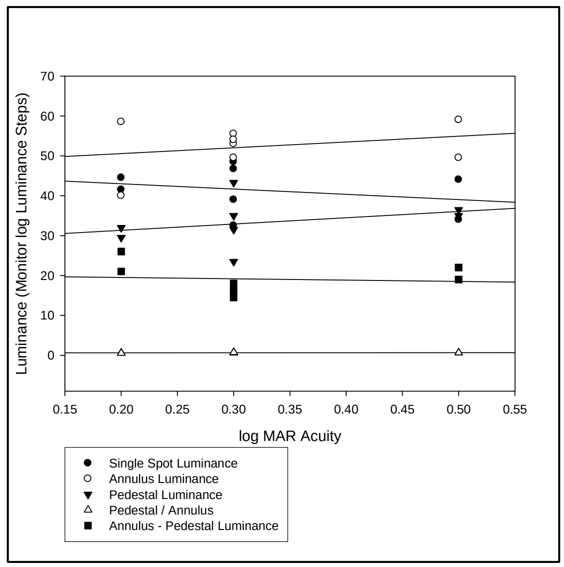


Correlations			
		Total RMS	Log MAR Acuity
Total RMS	Pearson Correlation	1	0.669
	Sig. (2-tailed)		0.07
	N	8	8

<u>Table C-5.</u>
Multiple Correlations to log MAR Acuity and Total RMS Data.

OD log MAR Acuity	OD Total RMS	Single Spot Luminance (monitor log luminance steps)	Annulus Luminance (monitor log luminance steps)	Pedestal Luminance (monitor log luminance steps)	Pedestal Luminance / Annulus Luminance	Annulus Luminance - Pedestal Luminance (Monitor log Luminance Steps	OD Total RMS
0.5	0.25	34.0	49.5	35.0	0.65	19.00	0.25
0.3	0.13	39.0	49.5	31.5	0.64	18.00	0.13
0.5	0.27	44.0	59	36.5	0.62	22.00	0.27
0.3	0.13	46.7	55.5	43.3	0.73	15.70	0.13
0.3	0.23	32.5	53	23.5	0.59	16.50	0.23
0.2	0.20	41.5	40	32.0	0.60	21.00	0.20
0.3	0.12	48.5	54	35.0	0.71	14.50	0.12
0.2	0.14	44.5	58.5	29.5	0.53	26.00	0.14





Correlations			
		RMS	Single Spot Luminance
RMS	Pearson Correlation	1	-0.575
	Sig. (2-tailed)		0.136
Single Spot Luminance	Pearson Correlation	-0.575	1
	Sig. (2-tailed)	0.136	

		RMS	Annulus Luminance
RMS	Pearson Correlation	1	-0.067
	Sig. (2-tailed)		0.874
Annulus Luminance	Pearson Correlation	-0.067	1
	Sig. (2-tailed)	0.874	

		RMS	Pedestal Luminance/ Annulus Luminance
RMS	Pearson Correlation	1	-0.308
	Sig. (2-tailed)		0.458
Pedestal Luminance/ Annulus Luminance	Pearson Correlation	-0.308	1
	Sig. (2-tailed)	0.458	

		log MAR Acuity	Pedestal Luminance
log MAR Acuity	Pearson Correlation	1	0.318
	Sig. (2-tailed)		0.442
Pedestal Luminance	Pearson Correlation	0.318	1
	Sig. (2-tailed)	0.442	
All the other correlations were < 0.300.			

Correlations			
		Total RMS	Pedestal Luminance - Pedestal Luminance
Total RMS	Pearson Correlation	1	0.240
	Sig. (2-tailed)		0.567
	N	8	8
Pedestal Luminance - Pedestal Luminance	Pearson Correlation	0.240	1
	Sig. (2-tailed)	0.567	
	N	8	8

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